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Contributions to successful trip recovery in younger and older adults

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Abstract

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Fall injuries lead to substantial health, economic and social costs. Fall risk increases with age, with tripping the main cause of falls. The aim of this thesis was to further understand the biomechanical contributions to successful trip recovery in younger and older adults.

A combined experimental and computer simulation approach was utilised. A 10-segment torque-driven model of trip recovery was developed and evaluated for the first phase of trip recovery. Ground contact was modelled with spring-damper systems. Hill-type torque generators generated segmental motion based on ramped activation functions. Trips were induced by applying a horizontal force at the toe of the swing limb.

A trip recovery experiment was also performed to address the research questions and obtain model input. A group of younger (n=8) and older (n=7) participants completed a trip recovery protocol in which kinematic, kinetic and EMG data were collected.

Main findings included that, during elevating strategy recoveries, younger adults were able to reduce the body's normalised forward angular momentum more with their recovery limb (0.011 m/s) than older adults (0.004 m/s), due primarily to increased knee joint moment. Older adults more often adopted a lowering strategy (79% of trips) than younger adults (41% of trips). Older adults exhibited a relatively high coactivation at the ankle of the recovery limb during elevating strategy recoveries. Younger adults showed varying muscle responses, while older adults had more consistent muscle activation responses. Younger adults used their arms more effectively during trip recovery to reverse the body's forward angular momentum (13%) than older adults (-3%). Older adults did not always increase their recovery step to provide more stability, as younger adults did. This is possibly due to their slow response and movement velocity.

It was suggested that during elevating strategy recoveries younger adults used an energy absorbing strategy, with absorption at the knee, whilst older adults used a pivoting strategy, which can be described by pendular motion with a rotational spring at the base.

Publications

Journal paper:

Roos, P.E., McGuigan, M.P., Kerwin, D.G. and Trewartha, G. (accepted for publication) The role of arm movement in early trip recovery in younger and older adults. *Gait and Posture.*

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Roos, P.E., McGuigan, M.P., Kerwin, D.G. and Trewartha, G. Development and evaluation of a computer simulation model of trip-recovery. *XIth International Symposium on Computer Simulation in Biomechanics*, 2007.

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Nomenclature

act1	time when torque generator is initially switched on
act2	ramping time from minimum to maximum activation
act3	initial activation level
act4	maximum activation level
act4b	maximum activation level just after the trip stimulus
act5	time when torque generator is first switched off
act6	time when torque generator is switched on again
act7	time when torque generator is finally turned off
AL	arm length
a_{max}	maximal activation
a_{min}	minimal activation
$Ankle_{Ydispini}$	displacements of the springs of the ankle at start of simulation
$Ankle_{Zdispini}$	position of the vertical springs of the ankle at ground contact
ARA	arm recovery amount
a_{trip}	coefficient for the reciprocal function for trip force
a_{trip}	activation value during trip and trip recovery
a_{walk}	activation value during walk
BW	body weight
BW*LL	body weight time lower limb length
$BW*LL^2$	body weight times lower limb length squared
$coact_{ankle}$	coactivation of the tibialis anterior and gastrocnemius
$coact_{hip}$	coactivation of rectus femoris and biceps femoris
$coact_{knee}$	coactivation of biceps femoris and rectus femoris
CM	centre of mass

CoP	centre of pressure
EMG	electromyogram
ERA	early recovery amount
exp_i	value of variable from experimental data
F_{lc}	force measured with the load cell
F_{lctr}	negative peak in F_{lc} just after start of the trip
F_{lctr}	negative peak in F_{lc} just after start of the trip
F_{trip}	horizontal force applied to the foot to induce a trip
$F_{tripmax}$	maximum horizontal trip force
F_x	medio-lateral horizontal ground reaction force
F_{xfc}	positive peak in F_x with ground contact
F_{xpo}	negative in F_x with push-off
F_y	anterior-posterior horizontal ground reaction force
F_{yexpi}	experimental anterior-posterior horizontal ground reaction force
F_{yfc}	negative peak in F_y on ground contact
$F_{ymaxexp}$	maximum experimental anterior-posterior horizontal ground reaction force
F_{ypo}	positive peak in F_y during push off
F_{ysimi}	simulated anterior-posterior horizontal ground reaction force
F_{ytc}	negative peak in F_y on contact with the tripping device
F_{zexpi}	experimental vertical ground reaction force
F_z	vertical ground reaction force
F_{zfc}	positive peak in F_z with ground contact
$F_{zmaxexp}$	maximum experimental vertical ground reaction force
F_{zpeak}	peak vertical ground reaction force
F_{zsimi}	simulated vertical ground reaction force

GA	medial gastrocnemius
GM	gluteus maximus
GRF	ground reaction force
k	spring constant
k_2	rate at which the torque drops off from the optimum angle
k_{ankle}	torsional stiffness of ankle joint
k_{hip}	torsional stiffness of the hip joint
k_{knee}	torsional stiffness of the knee joint
LL	lower limb length
LRA	late recovery amount
m	parameter that governs the rate at which the activation increases with angular velocity
$Met_{Ydispini}$	displacements of the springs of the met at start of simulation
$Met_{Zdispini}$	position of the vertical springs of the metatarsal at ground contact
p	smoothing parameter for Woltring's B-spline
$penalty_{exp}$	penalty given if model variable exceeds variable boundaries during model evaluation
$penalty_{FC}$	penalty given if model simulation failes to converge during model evaluation
$penalty_{ROM}$	penalty given if any of the joint angles exceeds its range of motion during model evaluation
r	damper constant
ref	reference value
RF	rectus femoris
RMS	root-mean-square
RMS_{angles}	RMS difference of experimental and simulated joint angles time histories

RMS_{CM}	RMS difference of the experimental and simulated CM position
RMS_{GRF}	RMS difference of the experimental and simulated ground reaction force during the first recovery step
sim_i	value of variable from simulation
SM	semimembranosis
Score	RMS difference between the simulated and experimental data
$Score_y$	score for the objective function used to define the horizontal spring and damper parameters
$Score_z$	score for the objective function used to define the vertical spring and damper parameters
T_0	maximum isometric torque
TA	tibialis anterior
T_{max}	maximum torque in the eccentric phase
$Toe_{Ydispini}$	displacements of the springs of the toe at start of simulation
$Toe_{Zdispini}$	position of the vertical springs of the toe at ground contact
t_{trip}	duration of contact with the tripping device
$t_{tripmax}$	time of the maximum trip force after the start of the trip
$t_{tripstart}$	time of trip stimulus
VL	vastus lateralis
$y_{exp}(t)$	experimentally obtained horizontal displacement of the spring-damper system
$y_{exp}dt(t)$	experimentally obtained horizontal displacement velocity of the spring-damper system
$z_{exp}(t)$	experimentally obtained vertical displacement of the spring-damper system
$z_{exp}dt(t)$	experimentally obtained vertical displacement velocity of the spring-damper system

$\theta_{ankleLdt}$	angular velocity of the left ankle joint
$\theta_{ankleRdt}$	angular velocity of the right ankle joint
θ_{cc}	contracile component angle
θ_{e}	extensor torque generator
θ_{ee}	elastic component angle
θ_{f}	flexor torque generator
θ_{hipLdt}	angular velocity of the left hip joint
θ_{hipRdt}	angular velocity of the right hip joint
$\theta_{kneeLdt}$	angular velocity of the left knee joint
$\theta_{kneeRdt}$	angular velocity of the right knee joint
θ_{opt}	optimum angle at which maximum torque occurs
ω_{1}	angular velocity at the point of inflection of the seven parameter function
ω_{c}	vertical asymptote of the Hill hyperbola
ω_{max}	maximum angular velocity at which joint torque can be produced

Chapter 1: Introduction

1.1. Context

The population in the developed world is ageing; in the year 2000 approximately 10% of the world's population was aged 60 or over, and this percentage is expected to rise to over 20% by 2050 (World Health Organization, 2003). As a consequence of this ageing trend, falls and fall related injuries will become an increasingly large and costly problem with extensive health, economic and social costs. Falls affect many people, with one-third of adults aged 65 and over falling at least once each year (Tinetti et al., 1988; Shephard, 1997). The severity and occurrence of falls increases with age (Campbell et al., 1981; Rogers et al., 2003a). Falls have substantial health costs as 29% of falls in people aged over 75 result in a serious injury, such as hip fractures, and falling is the sixth leading cause of death among older adults (Tinetti et al., 1988). In 40% of nursing home admissions falls are mentioned as a contributing factor (Tinetti et al., 1988). Falls appear to be more of a problem for women, since women generally outlive men (57% of the 65 year olds and over are female (Department of Trade and Industry, 2004)), but women also fall four times more often than men (Pavol et al., 1999b). Moreover, falls have psychological and social consequences. Older adults with a tendency to lose balance often have a reduced level of physical activity and a decreased ability to function in social roles (Rogers & Mille, 2003). Of people aged over 75 who had fallen 48% said they were afraid of falling and 28% said they were restricted in their normal daily activities due to their fear of falling (Tinetti et al., 1988). These health and social effects of falls lead to large financial consequences; in the UK the hospital and social costs following hip fractures alone amounts to more than £1.73 billion per year (National Osteoporosis Society, 2006).

Because of the extensive consequences of falls, effective fall-prevention programmes are important. The causes of falls are multiple and have been suggested to include reduced vision, vestibular impairment, reduced sensation, reduced static and dynamic balance, reduced walking velocity, poor mobility and gait disorders, reduced strength of the lower extremities, reduced reaction time, acute illness, chronic disease that affects sensory, neurological, cognitive and muscular function, cognitive impairment and polypharmacy (Rogers et al., 2003a).

Prevention of injury due to falling mainly works in the following three areas: 1) eliminating or reducing the hazards for loss of balance (e.g. check medication, correct prescription glasses, eliminate hazards in the home), 2) reducing the risk of a fall after loss of balance (e.g. by balance training or muscle strengthening) and 3) reducing the risk of injury when a fall occurs (e.g. by using hip protectors).

As falling is a multi-factorial problem, fall-prevention works best as a combination of approaches. This thesis is focussed on reducing the risk of an actual fall after loss of balance and targets primary fall-prevention, as people who have fallen once are likely to fall again (Tinetti et al., 1988). By targeting prevention of the first fall in relatively young older adults, subsequent falls at a later age could also be prevented.

Tripping is the most common cause of falling (Campbell et al., 1981; Tinetti et al., 1988; Tinetti & Speechley, 1989; Shephard, 1997), as 25-47% of falls in people aged over 65 are due to trips (Campbell et al., 1981; Tinetti et al., 1988). This thesis focussed on the biomechanics of trip recovery and investigated the changes in movement patterns that occur with age. The overarching goal was to work towards informing fall-prevention practices by identifying factors which contribute to improving trip recovery potential.

1.2. Key concepts

There are four concepts that are central to this thesis. A trip is considered to be:

> A trip is a perturbation of dynamic balance during normal walking gait. The linear movement of the body in the forward direction is obstructed at the base of support. This results in an angular movement of the body around the centre of pressure, with the body moving from an upright position, with its frontal aspect towards the ground.

And successful trip recovery is considered to be:

> Successful trip recovery is when dynamic balance is regained after a trip, i.e. the forward angular momentum is arrested and normal walking motion resumes. This balance has to be regained within one or several recovery steps.

Two terms used throughout this thesis are the 'initial stance limb' and the 'recovery limb'; they are used to differentiate between the limbs during trip recovery. Nomenclature of these limbs differs in the literature. As defined in this thesis, the limb that is prior to the object and initially supports the body is regarded to as the 'initial stance limb'. The limb that is lifted over the obstacle first and is put on the ground after the obstacle first is regarded as the 'recovery limb'.

1.3. Research aims

The overall research aim of this thesis was to further understand the biomechanical contributions to successful trip recovery in younger and older adults. Attention was focussed on a number of variables which should contribute to trip recovery success, such as joint moments of the recovery limb, response time, muscle sequencing, arm movements, range of motion and recovery step length.

The research aim was achieved by addressing the following hypothesis-driven research questions:

1: What is the contribution of the recovery limb in successful trip recovery in both younger and older adults?

The recovery limb is believed to further arrest the forward angular momentum of the body after the early reduction by the initial stance limb (Pijnappels et al., 2004).

Hypothesis 1: As maximum muscle force decreases with age, it was expected that younger adults would show larger joint moments in the recovery limb during trip recovery allowing a larger reduction of the forward angular momentum of their body than older adults.

2: How do muscle sequencing and coactivation influence successful trip recovery in both younger and older adults?

The timing of muscle activation is an important factor in coordinated movement (Hortobagyi & DeVita, 2006). Coactivation of agonist-antagonist muscle pairs increases the stiffness of joints (Hortobagyi & DeVita, 2000) and may therefore provide better stability, although it has efficiency costs. During trip recovery, balance is challenged and muscle timing will therefore be different from normal gait. It was expected that younger adults would show a more coordinated muscle activation sequencing during trip recovery, while older adults would use more coactivation and stiffen their joints to control movement.

Hypothesis 2: During trip recovery older adults will exhibit higher muscle coactivation than younger adults and muscle sequencing will differ between younger and older adults.

3: What is the contribution of arm movement to successful trip recovery in both younger and older adults?

The potential roles of arm movement in trip recovery are to assist by elevating the body centre of mass to give more time for proper placement of the recovery limb and by reducing the forward angular momentum of the body via relative backward rotations. Arm movements were expected to happen as part of the early response to a trip.

Hypothesis 3: Younger adults use their arms more effectively than older adults, due to an increased range of motion and generation of opposite angular momentum.

4: What is the difference in joint range of motion of the lower limb between younger and older adults, and how does this range of motion influence trip recovery?

Although older adults have in general a smaller possible range of motion than younger adults, it is not known whether the full range of motion is used during trip recovery.

Hypothesis 4: Older adults use a smaller range of motion of their lower limbs than younger adults and this limits trip recovery success.

5: How does the recovery step length vary in relation to trip recovery strategies in both younger and older adults?

Older adults show a larger lateral base of support than younger adults during normal gait (Hageman & Blanke, 1986). A larger base of support provides better stability and was therefore expected to increase trip recovery success. However during trip recovery, time for placement of the recovery limb is limited, which may results in a reduced recovery step length, and response time and movement velocity will be important.

Hypothesis 5: Older adults are not able to utilise a recovery step length as large as younger adults, and this limits trip recovery success.

In addressing the specific research questions it was hoped that certain important, more global, questions could be answered. For example, why does one person recover better from a trip than another person? How do younger and older adults differ from each other in trip recovery? Do older adults favour one particular trip recovery strategy that is different from younger adults? And if so, why?

1.4. Approach

The approach taken in this thesis was a combination of experimental and simulation modelling. This combination should provide certain advantages for answering the global research questions surrounding this topic. The simulation model better enables the investigation of the influence of individual variables and factors that are outside the current physical limits of the participants and can be manipulated. Experimental analyses provide data on real participants in realistic setting and also provide information to verify and evaluate the model.

A linked segment, torque driven computer model simulating trip recovery was developed to investigate the contributions to successful trip recovery. Inputs for this model were subject specific inertial values, torque generator activation time histories and initial motion data. Output was the resulting motion of the body segments, which contributed to overall body motion. A trip recovery experiment was performed with a group of younger (20-35 years) and a group of older women (65-75 years). Results of this experiment were used to compare trip recovery strategies of younger and older adults and for evaluation of the model. The success of trip recovery was quantified by the variable recovery amount (RA). This variable quantified the reduction of the forward angular momentum (around the centre of mass) during trip recovery.

For this thesis, the simulation model of trip recovery was developed and evaluated for the initial phase of trip recovery only. In future this model will be evaluated further and a sensitivity analysis will be performed. The model will be applied to identify and quantify the contributions to successful trip recovery. In addition to providing understanding of the biomechanics of trip recovery, it is hoped that the developed model will become a helpful tool in fall-prevention practices by providing subject-specific predictions of the outcome of therapy.

1.5. Thesis outline

The outline of this thesis is as follows.

Chapter 1: Introduction

This chapter has introduced the context of falls as a health issue for the population, the subject of fall-prevention, tripping and trip recovery, with some key concepts. It has posed the research aims with their research questions and hypotheses, and finally provided the research approach with its rationale.

Chapter 2: Literature review

This chapter reviews the relevant literature concerning ageing, falls, balance, fall-prevention, current fall-prevention research, experimental methods and approaches, and finally computer simulation models and modelling approaches.

Chapter 3: Trip recovery experiments

In this chapter the materials and methods of the trip recovery experiment are described, with details of the pilot testing that was performed prior to the experiment.

Chapter 4: Trip recovery simulation model

This chapter describes the development of a trip recovery computer simulation model and its rationale. It builds up from simple models to the final trip recovery model. Consideration is given to the estimation of model parameters and input parameters that need to be obtained experimentally.

Chapter 5: Model evaluation

This chapter describes the evaluation of the trip recovery model. It provides verification for the use of this model to simulate trip recovery.

Chapter 6: Contributions to trip recovery

This chapter describes the contributions to successful trip recovery that resulted from the outcomes of the trip recovery experiment. The chapter is divided into sections which align with the stated research questions and describe: the trial outcomes, joint moment analysis, muscle sequencing, arm movement, range of motion and recovery step length.

Chapter 7: Discussion and conclusions

This chapter discusses the outcomes of this research, discusses methodology and limitations and gives directions for future research.

Chapter 2: Review of Literature

2.1. Falls

2.1.1. Gait and balance

Two key concepts in the research field of falls and fall-prevention are balance and gait. Balance involves regulation of the static and dynamic relationships between the centre of mass of the body and the base of support. It requires the central nervous system to detect instability and for neural processing and the performance of the musculoskeletal system to generate the appropriate response (Maki et al., 2003). Balance can be divided into static balance (balance during standing) and dynamic balance (balance with a moving base of support). Static balance requires the position of the centre of mass to remain within the boundary of the base of support, otherwise corrective responses are required (Roberts, 1995).

There are several tests that measure balance control. Lajoie and Gallagher (2004) compared several of these tests and compared their outcomes between fallers and non-fallers. The tests they compared were: simple reaction time, the Berg balance scale, the Activities-specific Balance Confidence (ABC) scale and postural sway. The ABC scale consists of a subscale where subjects are asked to rate their confidence levels when asked to complete a number of activities. The Berg balance scale consists of 14 balance specific activities ranging from sit to stand to standing on one leg. These authors found that non-fallers have faster reaction times, higher scores on the Berg balance scale and the ABC scale and body sway at lower frequencies when compared with fallers (Lajoie & Gallagher, 2004).

Early responses to unpredictable perturbations of balance are too rapid to be voluntary movements and may even be too rapid for older adults to perform (Maki & McIlroy, 2006). Because of this, balancing reactions are often considered to be 'automatic'. However, there is growing evidence that high-level attentional and cognitive processing may be involved (Maki et al., 2003). Donelan et al. (2004) stated that dynamic stability depends on both passive musculoskeletal dynamics and active control by the central nervous system. It has been found that during perturbations of stance, early response reactions were not influenced by shifts in attention, whereas later responses 345 ms after EMG onset, were influenced by shifts in attention (Norrie et al., 2002). This suggests that early response reactions are automatic, while later responses are voluntary and can be influenced. This active control of balance suggests that balancing reactions, such as trip recovery, could be improved by appropriate training. This training would require a good knowledge and understanding of movement patterns during balancing reactions. To better understand balancing reactions the gait changes that occur with older age are described first with their effects on balance.

Features of functional and independent gait are the ability to support the upright body, maintain balance in the upright position and execute the stepping movement (Murray, 1967). The three main elements required for successful locomotion are: 1) the ability to generate and maintain fundamental locomotion patterns, 2) the ability to maintain basic dynamic stability between centre of mass and base of support, and 3) the ability of changing locomotor patterns in response to changes that threaten dynamic stability (Ferber et al., 2002).

After the age of 60, walking velocity decreases by 12-16% per decade (Smeesters et al., 2001; Rogers et al., 2003a). The slow walking velocity in older people has been believed to be caused by cautiousness and willingness to sacrifice walking velocity for increased accuracy (Payne & Isaacs, 1987), but also by mental and physical health status (Buchner et al., 1996). A slower walking velocity has been associated with a larger dynamic stability (England & Granata, 2007). However, Alexander (1982; 1992) showed that when an individual walks relatively slowly, correction of unwanted movement will take more time, leading to a greater chance to lose balance. This is due to the typical slow walking velocity of older adults demanding a larger control over forces on the ground and larger muscle control than a faster walking velocity would require (Alexander, 1992). This slower walking velocity is a possible underlying cause of the kinematic changes in gait that occur with older age; opinions however differ as to whether the change in walking velocity is a cause or effect of the kinematic changes in gait of older adults. Some authors state that changes in gait in older adults can be attributed to their typical slower walking velocity, as they disappear with a faster walking velocity (Spirduso, 1995). Not all changes in gait with older age can be attributed to a slower walking velocity as some changes in gait remain when walking with increased walking speed. These were changes such as increased anterior pelvic tilt, reduced ankle plantar flexion, reduced ankle power generation (Kerrigan et al., 1998), increased stride length variability and increased stride time variability, but not stride width variability (Grabiner et al., 2001). The fact that not all changes in gait with older age can be attributed to a typically slower walking velocity is supported by the finding that older adults are more likely to fall if they walk faster (Pavol et al., 1999a). If the changes in gait would disappear with a faster walking velocity, a smaller likelihood of falls would be expected. It can therefore be concluded that some but not all changes of gait that occur with older age are caused by a typically slower walking velocity.

The typical movement of an older person, compared with a younger persons, is slower, with a widened base of support, decreased step height and length (Hageman & Blanke, 1986; Payne & Isaacs, 1987; Spirduso, 1995), a greater hip extension, a reduced ankle plantarflexion and a greater tendency to out-toe (Payne & Isaacs, 1987). Contrary to this, vertical excursion of the centre of mass, lateral centre of mass excursion and pelvic rotation have been found to remain similar with age (Hageman & Blanke, 1986), although these authors also found walking velocity and stride width to be maintained. It is possible that these contradictory findings may be due to the fitness of the participants in the study, who were all healthy women aged over 60 and may not have shown all of the most common changes in gait that occur with age.

Joint moments and powers during gait become redistributed across the joints with increasing age, with more use of the hip extensors and less of the knee extensors and ankle plantar flexors (DeVita & Hortobagyi, 2000). During downwards stepping movements, maximum ankle moment values have also been found to be lower in older than in younger adults, while knee moment patterns are similar (Lark et al., 2003). Older adults cannot develop ankle moment as fast as younger adults and the ankle joint is less flexible than that of younger adults (Lark et al., 2003). Joint mobility, or range of motion becomes restricted with increasing age (Shephard, 1997). Irrespective of this, older adults walk with a greater range of motion at the hip and flex the hip joint more than younger adults (DeVita & Hortobagyi, 2000) and have been shown to use a greater percentage of their passive ankle and knee range of motion during downward stepping movements (Lark et al., 2004).

The fine control of body movements becomes limited with increasing age (Shephard, 1997). Postural sway usually increases with age (Payne & Isaacs, 1987; Spirduso, 1995; Sihvonen et al., 2005). Sihvonen et al. (2005) found a U-shaped dependency of body sway and age; the youngest (8 years old) and the oldest (93 years old) showed highest sway velocities when compared with people in the mid age range. Postural sway during quiet stance has been found to be correlated with postural control during mild perturbations (Hsiao-Wecksler et al., 2003), which suggest they might both be controlled by the same control mechanism. However, this contradicts the findings by Mackey and Robinovitch (2005) that postural steadiness during stance in older women did not correlate with their ability to recover balance. This suggests that postural sway is related to static balance only and not to dynamic balance.

Most of the gait changes in older age result in a more challenged balance and a higher incidence of falls. These are changes such as a slower walking velocity, reduced joint range of motion, slower joint moment generation, reduced step height, limited fine control of body movements and increased postural sway. Other changes in gait compensate for this challenged balance and increased risk of falling. These are changes such as a shortened step length, a widened base of support and a slower, more cautious gait. A slower walking velocity results in a more challenged balance when balance is lost; it however reduces the risk of actually losing balance.

These changes in gait and the higher occurrence of falls are caused by a number of physical and physiological changes in older adults which are described in the next section.

2.1.2. Physical changes with age in relation to falls

People often adapt a less active lifestyle with older age, although it is unknown whether this is a cause or effect of the physical changes that come with older age (Shephard, 1997).

Muscle strength changes with age; it is maintained after the age of 20-30, but rapidly declines after the age of 60 (Spirduso, 1995; Vandervoort et al., 2000; Rogers et al., 2003a). The decrease in muscle strength after the age of 60 can be partly attributed to a decrease in physical activity, but is also due to physical changes that come with age (Shephard, 1997). Muscle power becomes reduced with older age when scaled to body mass, as the percentage of body fat increases (Shephard, 1997). When muscle strength is adjusted for fat-free mass and muscle mass, age-related differences in muscle strength are present to a smaller extent (Frontera et al., 1991). The reduced muscle strength at older age influences mobility. The correlation between muscle strength and functional mobility is higher in women than in men (Samson et al., 2000). Functional mobility of older women has been found to be strongly influenced by ankle muscle power, and isometric ankle plantar flexor strength is a strong independent predictor for specific functional tasks (Suzuki et al., 2001).

Older adults cannot develop moment in the ankle joint as rapidly as younger adults, which is believed to be essential for fast recovery after loss of balance (Lark et al., 2003). Older adults need to use more muscles to maintain balance than younger people, because they bend more at the hip (Spirduso, 1995). Corrective responses to a loss of balance become slower with increasing age, and sometimes coactivation has a role with contractions of both antagonistic and agonistic muscle responses (Spirduso, 1995; Shephard, 1997). Muscle strength of older adults is lower, increasing the relative muscular demand during tasks such as stepping over obstacles. This results in a higher potential for muscle fatigue during locomotion and may place older adults at a higher risk of trips and falls (Hahn et al., 2005). Muscle activation is lower in older adults than in younger adults in a non-fatigued situation (Stackhouse et al., 2001), and in a fatigued situation this difference becomes more pronounced.

Older fallers differ from older non-fallers in several aspects. Fallers are significantly more asymmetric in power between legs (Skelton et al., 2002). Dorsi flexor strength of the ankle was found to be 7.5 times weaker in older fallers than in non-fallers (Spirduso, 1995). Fallers have smaller maximum ankle (17%) and knee (37%) moment than the non-fallers (Simoneau & Krebs, 2000). While no difference was found in whole-body angular momentum during normal gait between older fallers and non-fallers (Simoneau & Krebs, 2000), the consequences of these lower joint moments might be a smaller ability to control the body angular momentum during a fall. Low muscle strength is a risk for falls, although Skelton et al. (2002) proposed muscle power might be even more predictive. They showed that fallers were about 24% less powerful for their weight than non-fallers, when comparing the least powerful leg. Loss in back and leg strength increases the risk of falls in older people (Spirduso, 1995). Hausdorff et al. (1997) found that fallers and non-fallers differ in gait variability but not in velocity and gait cycle timing; variability was larger in fallers. There are several possible causes of the larger gait variability in fallers; these are central nervous system deficits, cardiovascular diseases and peripheral weakness (Hausdorff et al., 1997). It can therefore be suspected that it is not the gait variability itself, but the underlying causes of the gait variability that are the reason for the higher incidence of falls. Hausdorff et al. (1997) showed gait variability can be reduced by aerobic and progressive resistance muscle training.

Older adults have reduced ability to produce fine movements, resulting in older people having higher coactivation in demanding tasks (e.g. downward stepping) to create a higher leg stiffness (Hortobagyi & DeVita, 2000, 2006). Coactivation is when muscles that cause opposite joint actions are simultaneously active, serving to stabilise joints. This increased stiffness is also believed to be a compensation for reduced muscle strength and increased joint laxity; joints are stiffened and movement variability is reduced (Hortobagyi & DeVita, 2006). In general muscle coactivation is present in both younger and older adults when performing a novel task or a task that requires precision and joint stability (Patten & Kamcn, 2000). Task–specific training has been shown to reduce muscle coactivation, especially in older adults (Patten & Kamen, 2000).

The sensori-motor system deteriorates with older age (Samson et al., 2000), which may affect the detection of loss of balance and increase the risk of falling. Various aspects of vision decline with age (Payne & Isaacs, 1987). Vision has been shown to play an increased role in balance at older age (Chen et al., 2005; Poulain & Giraudet, 2005) and reduced vision has been found to be correlated with the occurrence of falls in older adults, particularly in women (Campbell et al., 1981).

Deterioration of various aspects of the sensori-motor system causes slower detection and reaction times with older age (Payne & Isaacs, 1987; Spirduso, 1995). This results in a slowed response to a perturbation of balance. Older people with a history of falling have a slower reaction time than those who do not have a history of falling (Grabiner & Jahnigen, 1992). Response time increases especially when generalisations must be made, a complex task must be undertaken (Payne & Isaacs, 1987), or several signals must be distinguished (Shephard, 1997). The ability of older adults to regain balance by taking rapid steps declines when there is only short time available for recovery (Thelen et al., 1997). Thelen et al. (1997) showed, using a sudden-release forward-leaning experiment with younger and older participants, that this decline seems to be mainly caused by the decrease in maximum velocity of the lower limbs, rather than by sensory or motor programming processes. This was as neither the younger nor the older participants increased their response time when less time was available for appropriate limb placement for recovery, while the younger adults recovered better in these situations.

Another physical change often occuring in people of older age that is a fall risk factor, is the high occurrence of foot problems, which are common in older adults and especially in women (Menz & Lord, 2001; Menz et al., 2005). Specifically, ankle flexibility, plantar tactile sensation and strength of toe plantar flexor muscles have an influence on balance (Menz et al., 2005). With ageing, cutaneous sensation of the plantar foot surface decreases, which is an increased risk for loss of balance (Maki & McIlroy, 1998). Foot motion and movement direction detection also decline with increasing age (Thelen et al., 1998).

There are also psychological factors that underlie the increased risk of falling in older adults. Fear of falling is believed to have a high impact on dynamic balance performance (Delbaere et al., 2005) and can therefore increase the risk of falling. People with a demonstrable fear of falling have been found to score worse on several balance tests than people without fear of falling (Maki et al., 1991). However, it is uncertain if the fear of falling is a cause or an effect of poor balance. Herman et al. (2005) showed that changes in the gait of people with fear of falling cannot simply be attributed to normal physiological or psychological consequences of ageing, but may be appropriate responses to unsteadiness and are the sign of an underlying pathology.

The physical changes that occur with older age described in this section are all underlying causes of the higher occurrence of falls in older adults. The most common cause of falling is tripping over an object (Campbell et al., 1981; Tinetti et al., 1988; Tinetti & Speechley, 1989). Recovery from a trip can be divided into two main strategies which are used by both younger and older adults. There are however some typical differences in trip recovery strategies between younger and older adults, which are described in the next section.

2.1.3. Trip recovery strategies

The two main trip recovery strategies used by both younger and older adults are the 'elevating strategy' and the 'lowering strategy' (Figure 2.1). Elevating strategies are used in response to perturbations in early-swing and lowering strategies to perturbations in late-swing, while perturbations in mid-swing can result in either an elevating or a lowering strategy recovery (Schillings et al., 2000). During a lowering strategy recovery a person attempts to rapidly lower the swing limb to the ground and to arrest the forward rotation of the stance limb (Eng et al., 1994; Schillings et al., 2000). The tripped foot is immediately lowered to the ground, prior to the obstacle. Then the tripped limb acts as the support limb as the other limb executes the initial recovery step across the obstacle (Pavol et al., 2001). This usually results in a flat foot or forefoot landing and a shortening of the step length (Eng et al., 1994; Schillings et al., 2000). During a lowering strategy, foot placement is actively controlled by rectus femoris and biceps femoris responses related to knee extension and deceleration of the forward sway (Schillings et al., 2000). Activation of tibialis anterior mostly preceded the main ipsilateral soleus responses, which is possibly to create a movement away from the obstacle (Schillings et al., 2000).

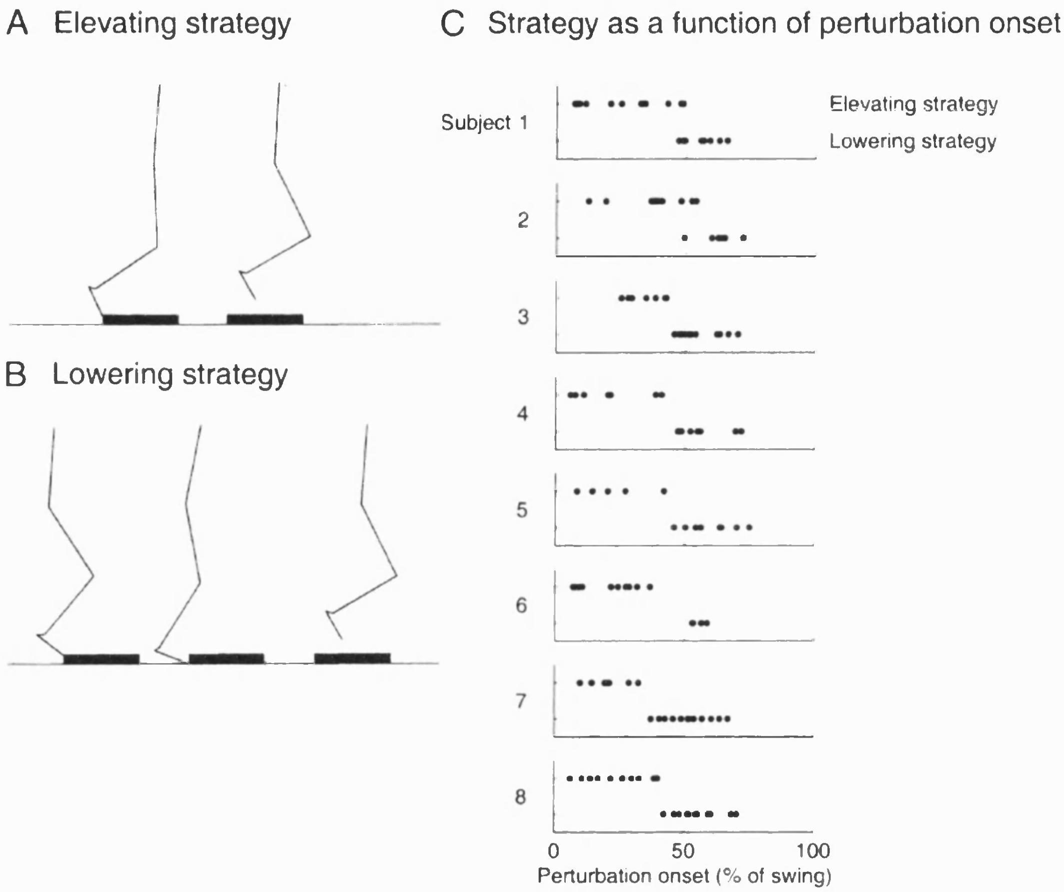


Figure 2.1 Response strategies. A: elevating strategy. B: lowering strategy. C: strategies performed as a function of perturbation onset in all subjects (% of swing indicates time of obstacle contact with respect to control swing duration) (adapted from Schillings et al., 2000).

During an elevating strategy (sometimes called reaching strategy), the tripped limb is used as the recovery limb as the foot is lifted over the obstacle in a continuation of the original step (Pavol et al., 2001). It requires a swing limb flexor component in addition to a stance limb extensor component. Multiple joints of the swing limb in addition to the stance limb and pelvis are used to elevate the body centre of mass. This provides extra time to extend the swing limb in preparation for landing (Eng et al., 1994). During an elevating recovery strategy the foot is directly lifted over the obstacle through extra knee flexion assisted by ipsilateral biceps femoris responses and ankle dorsi flexion assisted by tibialis anterior responses. Later, large rectus femoris activations induce knee extensions to place the foot on the ground (Schillings et al., 2000).

Elevating strategy recoveries have been found to be more energy efficient than lowering strategy recoveries in younger adults (Forner-Cordero et al., 2005). In lowering strategy recoveries the perturbed step is aborted and speed-loss must be compensated for in the recovery steps (Forner-Cordero et al., 2005). Lowering strategy recoveries take in general more strides to recover with than elevating strategy recoveries (Forner-Cordero et al., 2005), which indicates they are more challenging. It could be questioned whether energy efficiency is the governing criteria during trip recovery. An individual may choose to sacrifice energy cost for an appropriate recovery which prevents injury.

Older adults have been found to more often adopt a lowering strategy recovery than younger adults (Pijnappels et al., 2005a), although the reason for this has not been established. Older adults prefer to adopt a strategy that has been found to be less efficient energy-wise in younger adults, although it is unknown whether this strategy is also less efficient in older adults. Older adults might be unable to use an elevating strategy in some situations (e.g. due to reduced muscle strength or response time), or a lowering strategy might be more effective for older than for younger adults.

Falls during elevating and lowering strategy recoveries differ from each other and have different causes. During-step falls (early in recovery) in lowering strategy recoveries were associated with a faster walking speed at the time of the trip and delayed support limb loading. After-step falls (later in recovery) in lowering strategy recoveries were associated with a more anterior head-arms-torso centre of mass at the time of the trip, followed by excessive lumbar flexion and buckling of the recovery limb (Eng et al., 1994; Pavol et al., 2001). The elevating strategy fall was associated with a faster walking speed followed by excessive lumbar flexion (Pavol et al., 2001).

Both younger and older adults perform elevating and lowering strategies. There are however some typical differences in their trip recovery movement patterns. The characteristics of the first step of recovery from older adults are similar to those of younger adults up to the time of recovery foot-contact (Maki et al., 2003). Older adults are more likely to recover from a trip or perturbation of balance with multiple steps than with single steps (McIlroy & Maki, 1996; Mille et al., 2005) and take smaller steps than younger adults (Rogers & Mille, 2003). Balance impaired older adults require even more additional recovery steps after perturbations of balance than non-impaired older adults (Schulz et al., 2005). The need for multiple recovery steps in younger adults is normally larger during lowering than during elevating strategy recoveries, and can be attributed to the larger initial step-length and higher hip moments during elevating strategy recoveries (Forner-Cordero et al., 2004). The need for additional steps in trip recovery in older adults could be caused by their lower muscle strength, slower response time, or trip recovery technique.

Both younger and older active adults (runners) have been found to recover less often from a trip with multiple steps than inactive adults (Karamanidis & Arampatzis, 2007). This has mainly been attributed to increased motor skills (e.g. the ability to recover balance with a single-step after a fall) as the active and inactive adults had similar leg-extensor muscle tendon unit capacities (e.g. muscle strength, tendon stiffness, and muscle morphology) (Karamanidis & Arampatzis, 2007). This indicates that running increases motor skills and can partly compensate for age-related degeneration of leg-extensor muscle strength and tendon stiffness effects on regaining balance after forward falls (Karamanidis & Arampatzis, 2007).

Another difference in trip recovery between younger and older adults is that older adults remain in a flat-foot position for a long period of time before rising up on the ball of their foot after perturbation of balance (Lark et al., 2003). This is possibly to maintain a larger base of support for as long as possible. When crossing obstacles, healthy older adults have been found to have a more conservative strategy than younger adults (Hahn & Chou, 2004). They showed a slower crossing speed, shorter step length and step width and had an increased risk of obstacle contact (Hahn & Chou, 2004). Toe-clearance was smaller in older adults with a low functional level than for adults with a high functional level when stepping over an obstacle (Brunt et al., 2005). Trips over an obstacle can be expected to occur more often with the trailing than with the leading foot, because of the closer proximity of this foot to the obstacle (Chou & Draganich, 1997). Toe-stubbing led to more dangerous loss of balance than heel-stubbing (Murray, 1967). The gluteus medius, vastus lateralis and gastrocnemius are more challenged (higher activation levels) in older than in younger adults when stepping over obstacles (Hahn et al., 2005).

Older adults with balance disorders have been found to have larger and faster lateral centre of mass movement and medio-lateral swing foot motion when stepping over obstacles, while other gait variables, such as velocity, stride time and stride width were similar to those of healthy older adults (Chou et al., 2003). Step width, average trunk flexion, and the phase of the gait in which the trip occurred does not increase the risk of falling (Pavol et al., 1999a).

Older adults are more likely to grasp for support to recover from a trip than younger adults (Maki et al., 2003). This is only when their hands are free and holding a cane can reduce the grasping movement of the arm which can be important in arresting a possible fall (Bateni et al., 2004). Activation of the muscles in the shoulder that cause arm movement occurs very fast after perturbation of stance, and is similar in time to automatic responses (McIlroy & Maki, 1995). Falls in older adults are often in a forward direction; as a result upper extremities injuries are very common, presumably in protecting the head and torso (DeGoede & Ashton-Miller, 2003). Some studies that looked at arm movements during perturbation of stance suggested that arm movements may play a role in trip recovery. Older adults were more inclined to use their arms when perturbed from stance and stimulated not to use their arms by holding a rod behind their back (Maki et al., 2000). When arm movement was not prohibited older adults showed smaller arm movement after perturbation of stance than younger adults, and the movements were in opposite direction; older adults moved their arms in the same direction as the fall, while the arms of younger adults counter acted the fall movement (Allum et al., 2002). This may indicate older adults have a more protective arm movement while younger adults have a more fall preventive arm movement counter acting the forward angular momentum caused by the loss of balance. Older adults were more likely than younger adults to grasp for support with their arms in platform perturbation studies, although they were less able to make these arm movements (Maki & McIlroy, 2006). The role of arm movements in trip recovery has to date however not been investigated.

2.1.4. Fall-prevention

The high occurrence of falls in older adults makes fall-prevention essential. As tripping is the main cause of falling, investigating the biomechanics of trip recovery in older and younger adults can provide important information for fall-prevention practices. To understand the needs of and identify the gaps within current fall-prevention research, the next section discusses current fall-prevention research and practice.

Fall-prevention is a complicated area, mainly due to two reasons: clinicians are most experienced with discrete diseases while falling is a multi-factorial condition, and some components of fall-prevention require trade-offs, weighing up the risks against the benefits (Tinetti, 2003). Tinetti and Speechley (1989) stated that "the goal of preventive strategies should be to minimise the risk of falling without compromising the mobility and functional independence of the elderly".

Several fall-prevention programmes, such as professionally supervised balance and gait training, muscle-strengthening exercise, gradual discontinuation of psychotropic medications, and modification of hazards in the home after hospital discharge, have proven to be effective (Tinetti, 2003). Feder et al. (2000) said exercise alone did not reduce the rate of falls and multifaceted interventions were needed, while others did find a reduction in falls or an improvement of motor functions after physical exercise programmes with older adults. Rijken et al. (2005) developed a short term (5-week) exercise program with balance, walking and fall technique exercises derived from martial arts, and decreased the number of falls by 46% in their exercise group. Nnodim et al. (2006) found that with a combined balance and step training programme better improvements in balance and stepping measures were obtained than with Tai Chi training. Skelton and Beyer (2003) reviewed exercise interventions for fall-prevention and found that different age groups benefited from different types of intervention; those over 80 who fell frequently and injured easily benefited most from supervised home-based exercise programs, while younger, community-dwelling fallers benefited more from multifactorial group interventions targeting balance, strength, power, gait, endurance, flexibility, co-ordination and reaction.

Strength training has also been shown to be effective in fall-prevention. High resistance training and agility training have been shown to be more effective than a stretching program in reducing fall risk in older women with low bone mass (Lui-Ambrose et al., 2004). Lord et al. (2003) investigated the effect of weight bearing group exercise against a control of flexibility and relaxation classes on the occurrence of falls in older adults. They found there were 22% fewer falls during the trial in the weight bearing group than in the control group, and a total of 31% fewer falls in the subjects who had fallen in the past year and followed the weight bearing group exercise programme. These findings suggest that strength training and weight bearing exercises are more effective in fall-prevention than flexibility training and stretching.

Step initiation timing is another fall risk factor that can be improved by step training (Rogers et al., 2003b). A three-week step training programme improved step initiation timing of older adults by 17% (Rogers et al., 2003b). Best results were obtained for induced step training, where stepping was induced by waist pulls, rather than voluntary step training, where people were asked to step after an auditory cue (Rogers et al., 2003b).

Accurate control of force, which is reduced in older adults and important for fall-prevention, can be improved by training. Patten and Kamen (2000) showed that a six-week force modulation training program can lead to improvements for both younger and older adults in force accuracy, and to an increase in the maximal voluntary force for younger adults only.

Body sway, which by some researchers has been associated with an increased fall risk, and balance can be improved by randomly vibrating insoles, producing an inout noise signal, especially in older adults (Priplata et al., 2003). The stimulation level of these insoles was set just below the sensory threshold of the subjects and is believed to enhance sensory feedback. These insoles have not been tested in dynamic situations.

Because falling is such a multi-factorial problem a personalised intervention will get the best results for fall-prevention therapy. A better insight in the movement patterns of trip recovery could possibly enrich current fall-prevention practices. A tool to help clinicians chose the right approach, such as a computer simulation model that showed the outcome of the therapy, would be useful to choose the right personalised therapy. The following section will discus research in trip recovery, and highlight the insights in trip recovery that are still missing.

2.2. Research into trip recovery

Considerable research has already been done into trip recovery strategies. However substantial gaps still exist in the knowledge base of how to reduce falls from trips. The following section discusses experimental research in trip recovery to identify the best experimental methods available and to highlight gaps in the findings of trip recovery research which could be addressed to provide valuable information for fall-prevention.

2.2.1. Experimental approaches

The choice of the participant group is an important aspect of trip recovery experiments. Trip recovery movement patterns differ between younger and older adults. Insight into trip recovery of older adults can only be fully obtained by experiments with older participants. Selection of an older participant group comes with high ethical requirements, since older adults may be frail and have an increased risk of injury. Consequently only a small number of research groups have used older participants in their trip experiments (Wu, 1998; Pavol et al., 1999b; Rogers et al., 2001; Wojcik et al., 2001; Pijnappels et al., 2005a; Troy & Grabiner, 2005). Mori et al. (2006) used an animal model instead of human subjects in their study of obstacle clearance, namely the Japanese monkey. They stated the strategies in the monkey, who was genetically quadrupedal and taught to walk bipedally on a treadmill for this study, were similar to those found in humans (Mori et al., 2006). It can however be questioned whether the fine movement patterns of the monkey resemble those of humans and whether age-related effects on movement patterns can be studied with this animal model.

Studies investigating perturbations of balance used various ways to induce trips or perturb balance from stance. Static balance was often perturbed by producing sudden movements of the support surface (Romick-Allen & Schultz, 1988; Wu, 1998; Troy & Grabiner, 2005; Maki & McIlroy, 2006), or by sudden release, or pulling experiments with ropes to the waist (Thelen et al., 1997; Wojcik et al., 2001; Mackey & Robinovitch, 2005; Schulz et al., 2005) or lower leg (Smeesters et al., 2001). Some researchers simulated trips by suddenly arresting the movement of the lower limb with a rope during walking (Forner-Cordero et al., 2005). Trips closest to "natural trips" were induced by placing mechanical objects in the walkway, either by dropping (Schillings et al., 2000), rising from below the floor (Grabiner et al., 1993; Pavol et al., 1999b; Burg van der et al., 2005; Pijnappels et al., 2005c) or rotating upwards (Eng et al., 1994). These trip stimuli were either induced manually (Grabiner et al., 1993; Pavol et al., 1999b), triggered by the subject's weight on a force plate (Eng et al., 1994), or online kinematics were used to calculate the place and time where an obstacle had to appear to induce a trip in mid-swing (Pijnappels et al., 2005c).

The kinematics of older adults have been found to differ when tripping over a real obstacle and when using surrogate tasks (sudden release and rapid platform displacement) to induce a tripping movement (Troy & Grabiner, 2005). Step height, step length and peak horizontal velocity of the leading and trailing limbs were significantly different between these situations (Troy & Grabiner, 2005). It can therefore be concluded that when studying movement patterns during trip recovery an experimental protocol where trips are induced by obstacles will give the best results, i.e. trips that are closest to "natural trips". It is also necessary that the participants are unable to hear or see this object appearing. This can be achieved by playing music and using glasses that obscure the lower half of vision (Pijnappels et al., 2001).

As failed trip-recoveries may lead to falls, safety measures are required. In the majority of trip experiments subjects were secured in a safety harness to prevent possible falls (Romick-Allen & Schultz, 1988; Wu, 1998; Pavol et al., 1999b; Schillings et al., 2000; Rogers et al., 2001; Wojcik et al., 2001; Forner-Cordero et al., 2005; Pijnappels et al., 2005c). These safety harnesses were generally suspended on an overhead rail or track. Pijnappels et al. (2001) suspended the harness with elastic ropes or a visco-elastic brake to minimise the impact on the subjects. Smeesters et al. (2001) bordered their walkway on each side with soft mats and added an additional mat in case of a forward or backward fall, next to this they padded the walkway with firm foam. Some used extra safety measures besides a safety harness, such as a handrail to grasp when needed (Schillings et al., 2000; Ferber et al., 2002), or cotton to protect the toes (Schillings et al., 2000). A minority of authors performing trip recovery experiments did not describe any safety measures taken in their experiments at all (Grabiner et al., 1993; Eng et al., 1994, 1997).

It has to be considered that the participants in trip trials are informed prior to the trials that trips will be induced and that this might alter their gait. Pijnappels et al. (2001) investigated whether gait was indeed altered by forewarning of possible trips. They found forewarning of possible trips caused small changes in several spatial parameters (such as step width and foot clearance), but the temporal parameters (such as walking velocity, step frequency, duration of the stride cycle, stance, swing and double support time and step length) remained similar. Similar research by Forner-Cordero et al. (2003) confirmed these results. Pijnappels et al. (2006) also investigated whether muscle activity was altered after after a trip occurred due to forewarning. They found increased co-activation after tripping in hamstring, quadriceps and tibialis anterior muscles of younger adults, indicating that younger adults attempt to stiffen their joints when a trip can be expected, while the older adults only had increased muscle activity in tibialis anterior and soleus. The increases in muscle activity were small and Pijnappels et al. (2006) concluded they were not large enough to threaten validity of experiments with more than one trip trial. Forewarning of tripping can therefore be expected not to change the probability of tripping or the tripping response itself.

When comparing results from different trip recovery studies, external factors that may influence the biomechanics, such as obstacle height, have to be considered. Experiments where subjects stepped over obstacles showed increased adduction and internal rotation moment of the hip joint during early stance, increased internal rotation moment at the knee joint during late stance and increased dorsi flexion moment at the ankle during late stance with increasing obstacle height (Chou & Draganich, 1998). The amount of joint work required to step over an obstacle increases linearly with obstacle height (Chou et al., 1997). Older adults linearly increased leading toe clearance with obstacle height, changing fewer joint angular components than younger adults and maintaining stability with a minimum control effort (Lu et al., 2006). Centre of mass movement in anterior-posterior and vertical directions, vertical velocity of the centre of mass and anterior-posterior distance between the centre of mass and centre of pressure have also been found to increase when stepping over obstacles of increased height, while medio-lateral displacement of the centre of mass remained similar (Chou et al., 2001). It can be concluded that to be able to compare results with contemporary research in this field and to induce trips as close to "natural trips" as possible the most appropriate way to induce trips is by using obstacles in the walkway. As some of the research questions posed in chapter 1 relate to differences between younger and older adults this study used both younger and older participants.

2.2.2. Relevant outcomes of trip recovery research

Considerable research has been done into trip recovery; therefore this section focuses only on the research relevant to the research questions posed in Chapter 1. The studies by Pijnappels et al. (2004; 2005a; 2005c; 2005b) are similar to those described in this thesis. They have answered several questions regarding trip recovery, however some were left unanswered. They used a group of older non-fallers, a group of older fallers and a group of younger participants and obstructed normal gait. They looked at the contribution of the support limb (initial stance limb) in trip recovery and at muscle sequencing and timing. They measured kinematics and ground reaction forces at the support limb, together with

EMG of muscles in the lower limbs. They calculated the external moment, which equals the rate of change in the angular momentum of the body.

This series of studies showed that the support limb plays a role in trip recovery by providing time and clearance for proper positioning of the recovery limb. Almost all young participants were able to restrain the forward angular momentum of the body during push-off by the support limb. None of the older participants was able to fully reduce the angular momentum during push-off, but the older participants without a history of falls were able to stop a further increase in angular momentum. The push off reaction to restrain the forward angular momentum of the body was generated by a large ankle plantar flexion, a large knee flexion and a large hip extension moment. The onset of knee joint moment generation was slightly later in the older than in the younger adults, the rate of change of joint moment generation was in all joints lower for the older than for the younger adults.

The EMG data from the Pijnappels et al. (2005b) studies showed similar amplitudes for the younger and the older adults. However, the rate of increase of muscle activation during recovery was reduced in the older adults. Both the younger and the older adults showed rapid responses (60-80 ms) in their hamstrings and triceps surae muscles, followed by responses (90-130 ms) in the quadriceps muscles. The sequence of muscle activation was the same in the younger and the older adult group. The support limb responses were recovery strategy dependent. In the elevating strategy the hamstrings and triceps surae muscles stayed activated, leading to a prolongation of the push-off while the obstructed swing limb was placed forward, and the vastus lateralis muscle was activated which resulted in knee extension. In the lowering strategy, the hamstrings and triceps surae muscles were deactivated and the rectus femoris muscle was activated, resulting in knee extension. Furthermore, a late tibialis anterior activity was seen in the lowering strategy.

They found the younger adults achieved an aerial phase during recoveries, which the older adults did not. Some older adults performed a lowering strategy recovery when an elevating strategy would be expected. The older fallers improved their recovery success over trials.

Madigan and Lloyd (2005) found joint moment patterns similar to those found by Pijnappels et al. (2005a) in younger and older adults during balance recovery from a forward-lean sudden release experiment. The older adults however showed smaller peak knee extensor moments during the support phase of recovery, and tended towards larger peak extensor moments at the ankle and hip. The differences in the moments found in these studies can probably be explained by the difference between the experimental protocol, as Pijnappels et al. (2005a) looked at trip responses and Madigan and Lloyd (2005) looked at balance recovery after sudden release.

There are still some areas of trip recovery that have not been investigated and could provide essential information for fall-prevention. The following questions are still left unanswered: what is the contribution of the recovery limb to successful trip recovery?; what is the contribution of arm movement to successful trip recovery?; why do older adults recover more often with the less efficient lowering strategy than younger adults?; how large is the inter-individual variation in trip recovery and what are its underlying causes?

Not all these questions can be answered by experiments alone. Using a combination of an empirical and computer simulation modelling approach to answer these questions would have some advantages, as described in chapter 1. The following section describes some computer simulation models that investigate trip recovery responses.

2.3. Modelling trip recovery

A small number of a few research groups have used computer simulation modelling approaches to investigate trip recovery. These simulation models have typically been relatively simple in comparison to those applied in other areas of biomechanics. The majority of trip recovery simulation models are inverted pendulum models.

A simple inverted single pendulum model was developed by van den Bogert et al. (2002) to determine the relative importance of initial body position, walking velocity and response time to preventing a fall after a trip. The model simulated a trip by representing the body as a single rigid rod, fixed at the base that rotated as an inverted pendulum. The movement of the body stopped at the response time, and at this moment a hypothetical foot stopped the movement. It was assumed that the angular velocity of the body just before the trip was equal to zero. Simulations with this model showed that body tilt angle was a perfect predictor of a successful recovery step and that a faster response time was more important than a slower walking velocity for successful recovery.

Hsiao et al. (1999) developed a simple pendulum-spring model to investigate how step length and step contact time influence the effort and feasibility of balance recovery by stepping. The model consisted of an inverted pendulum, representing the head, torso, upper extremities and pivot leg, a mass-less linear spring, representing the stepping leg and a rotational spring, representing the restraining action of the stance limb. The simulations consisted of two phases, the pre-contact and the contact phase. In the pre-contact phase again the body was represented as a rigid rod, fixed at the base, and rotating as an inverted pendulum. At the base a rotational spring was added, simulating the net effect on the body's downward rotation of opposite lower extremity moments. In the contact phase the pivot leg contacted the ground and the linear leg spring was compressed. Recovery effort was defined as the peak contact force and energy absorbed in the stepping leg during step contact. Energy expenditure during step initiation was not considered. Successful recovery was signified by the occurrence of a negative angular velocity at a value smaller than 90° for the angle between the vertical and the body lean axis. The model predicted that successful balance recovery by stepping was controlled by coupling between step length, step time and leg strength. An important limitation of the model was that it did not simulate swing-phase dynamics.

A more complex model was developed by Forner-Cordero (2004) to simulate recovery from a stumble. This model aimed to control trunk motion as an inverted pendulum by appropriate foot placement during double stance. The model consisted of three links, the trunk and two legs. Trunk motion was modelled by inverted pendulum motion around the hip joint and a torque was applied here. The moment at the hip was dependent on the centre of pressure of both feet. Simulations with this model showed that steps of high velocity were required for successful recovery from a stumble.

To obtain more detailed insight into the kinetics of trip recovery and investigate the influence of multiple variables on trip recovery a more complex model is required than the described inverted pendulum models. A linked-segment model with rigid segments representing the limbs of the body would be able to provide more detailed insight into trip recovery repsonses. Joint moments represent summed forces at the joints and enable investigation of these forces during trip recovery. To date no linked-segment models have been published that simulate trip recovery. To simulate the recovery steps taken during trip recovery a bipedal model is required.

A simple bipedal model was developed by Alexander (1992) to model human locomotion. Each leg had a telescopic actuator that could exert force and make it lengthen and shorten, representing the muscles that change the length of real legs by flexing and extending the knee and ankle joints. A compression spring aligned with the long axis of the leg represented the elastic compliance conferred on real legs by the properties of tendons and ligaments. Torque actuators at the hip represented the muscles that flex and extend real hip joints. The knee and ankle joints were however not represented in this model. In trip recovery precise and balanced movements at these joints are essential. Rigid limbs, without ankle and knee joints, would make it difficult to step over an obstacle. Bothner and Jensen (2001) developed a bipedal model with ankle, knee and hip joints to simulate balance control during sudden support surface movements. This model consisted of four segments (foot, shank, thigh, and head-arms-trunk). This model was used to calculate moments acting at the ankle, knee and hip joints. In trip recovery steps are taken to regain balance, in contrast to the platform perturbations modelled by Bothner and Jensen (2001). These recovery steps require an accurate simulation of the ground reaction forces. Examples of how these ground reaction forces can be modelled are given in section 2.3.2.

A model can be personalised for different subjects by using subject-specific anthropometrics and inertia data as input to the model. There are several methods to calculate inertia parameters, some more accurate than others. These methods include cadaver-based measurements, mass scanning and mathematical models. Mathematical models are in general the most accurate, but they are generally the most time consuming. Dempster (1955) was one of the first to present cadaver-based data. Durkin and Dowling (2003) and Zatsiorsky and Seluyanov (1980; 1983) both used scanning methods to calculate body parameters; dual energy x-ray absorptiometry (DEXA) and gamma-scanner respectively. Several mathematical models have been developed. Hatze (1980) created a method that requires 242 anthropometric measurements, which can be carried out in 80 minutes. This model differentiates for male and female subjects.

Another mathematical inertia model frequently used in biomechanics to obtain individual specific segment inertia parameters was developed by Yeadon (1990). The method requires 95 anthropometric measurements which makes it time consuming, and was initially developed for application to sporting populations. A method to define inertial parameters (mass and centre of mass) optimised for use on older adults was developed by Pavol et al. (2002). It uses a relatively small number of anthropometric measurements (32) all of which are easily obtainable. The method uses a combination of the following approaches: body mass and segment length proportions, linear and non-linear regression equations and a mathematical model of the trunk (from Yeadon (1990)).

Two important choices in the development of a linked segment simulation model of trip recovery are the choice of the muscle actuators and the choice of the method to model ground reaction forces. The actuators will control the movement and the ground reaction forces will support and stabilise the body during trip recovery; these issues will be described in detail in the following sections.

Another important part of model development is model evaluation, as it establishes in a quantitative manner the level of accuracy that may be expected from a model (Yeadon & King, 2002). Model evaluation is described in more detail in chapter 5.

2.3.1. Choice of actuators

The two main approaches to analyse the kinetics of movement are: an inverse dynamics approach in which the internal forces and moments are reconstructed from the movements and known external forces, and a forward (direct) dynamics approach in which the motion is calculated from known internal forces, moments and resulting reaction forces (Runge et al., 1995; Otten, 2003). The main disadvantage of an inverse dynamics is often considered to be the error introduced via the amplification of noise by numerically differentiating position data twice (Koopman et al., 1995; Anderson & Pandy, 2001). Inverse dynamics solutions are however useful initial guesses for forward dynamics optimisation algorithms (Rasmussen et al., 2001; Otten, 2003). Forward dynamics simulations allow for investigation into the influence of varying the magnitude and timing of internal forces on resulting motion, allowing investigation of the effect of force and coordination on trip recovery responses.

Muscle action can be simulated by modelling individual muscles and their activation profiles, or by joint moments, representing the resultant action of muscle groups on a joint. Individual representation of muscles however has the following two difficulties: the indeterminate problem (when using inverse dynamics) and establishing the moment arms of the muscles over the complete range of motion of a joint (Rasmussen et al., 2002b). The human body is statically indeterminate: there are not enough equilibrium equations available to uniquely determine the muscle forces and joint reactions in each position (Rasmussen et al., 2001; Rasmussen et al., 2002a). Vaughan (1995; 1996) gave three main strategies to overcome this indeterminate problem: 1) reduce the number of unknowns (e.g. assume that only one muscle exerts a force), 2) use a mathematical optimisation theory, unknown forces will be treated as design variables, the task is to find those that minimise a cost function, 3) reduce all muscle, bone and ligament forces to a single (vector) resultant force and moment. Thelen et al. (2003) used the second strategy to overcome the indeterminate problem in their model with 30 muscles, activated by muscle activation algorithms. They developed the "computed muscle algorithm", which uses feed-forward and feedback control to drive the kinematic trajectory. The third strategy is the most common; muscles are grouped together by the joints they act on and are represented as joint moments.

The most important factors for the force development of a muscle are its contraction velocity and activation. The Hill-type model is the simplest model that meets these criteria (Gerritsen et al., 1995; Wagner & Blickhan, 1999). Hill-type models are often used to simulate muscle forces and joint moments. Hill described the force-velocity relationship of muscle during shortening as a rectangular hyperbola (Hill, 1938):

Equation 2.1 $(P+a)(v+b)=(P_0+a)b$,

where P is the load applied, v the shortening velocity, and $P = -a$ and $v = -b$ are the asymptotes of the hyperbola.

Joint moments can be modelled similarly by functions based on the peak moment, and the moment, joint angle and angular velocity relationship to provide an angular equivalent of the Hill approach. King and Yeadon (2002) used this Hill-type muscle model, which is described in more detail in chapter 3. Moment generators can simulate muscle force more naturally by using ramped activation functions to represent delay from initial stimulation to development of force; when activated the moment ramps up from an initial to a maximum value and when deactivated the moment ramps down to zero. King and Yeadon (2004; 2005) developed such a ramped activation model to simulate Hecht vault and tumbling in gymnastics. This activation model is described in more detail in chapter 4.

2.3.2. Modelling the ground reaction forces

Modelling ground contact with its ground reaction forces is a complicated and important part of the simulation of human movement. There are several ways in which ground contact can be modelled, a selection of these are described in this section.

As the moment of impact is a complicated part of modelling ground contact, this part can be avoided in simulation models. Mu and Wu (2000b; 2004) did this in their five link model of walking. They say the motion during the double support phase is difficult to control and approached this by developing a sliding mode controller for motion regulation during the double support phase. It was therefore assumed that the tips of both limbs were fixed to the ground and impact was assumed to be perfectly plastic.

As impact is not perfectly plastic, a better way to simulate ground contact is by calculating the ground reaction forces using spring and damper equations. This method is often used in biomechanics and it requires information on the elastic and damping properties of the floor-foot combination and the length and velocity of the springs and dampers at initial contact. The spring and damper equations used to simulate ground reaction forces differ in complexity and in the number and placement of the spring-damper systems. For example, Andrews and Dowling (2000a; 2002) used a fourth order vertical mass-spring-damper to model heel impact during landing after a sudden drop, while Wilson et al. (2006) used two second order horizontal and vertical spring-damper systems to model foot contact during takeoff of running jumps. Gilchrist et al. (1997) even used nine contact elements to model foot contact during walking.

Gerritsen and van den Bogert (1995) and Gilchrist et al. (1997) used, a combination of visco-elastic elements to model the ground reaction forces and Coulomb friction or shear forces to model the friction between the foot and the ground. Modelling friction between the foot and the ground with separate equations makes simulation of ground contact however unnecessary complex, as friction can be incorporated into the horizontal spring-damper systems that calculate the ground reaction forces.

2.4. Conclusions

Some of the physical changes that occur with older age increase the risk of falling in older adults. Tripping is a common cause of falls in older adults. Younger and older adults use similar strategies to recover from tripping, although younger adults are able to recover more successfully. Research into trip recovery and fall-prevention have shown some of the reasons why younger adults are able to recover from a trip more successfully than older adults. However, there are some gaps that still require investigation: the contribution of the recovery limb to successful trip recovery; the contribution of arm movement to successful trip recovery; the reason why older adults more often use the less efficient lowering strategy than younger adults; and the amount and underlying causes of inter-individual variation in trip recovery. The use of a combined empirical and computer simulation modelling approach to answer these questions would have some advantages. Several simulation models have already been developed to investigate trip recovery, but these have all been relatively simple models. A more complicated linked segment model with joint torques would be able to investigate trip recovery responses in more detail and possibly provide new insights.

Chapter 3: Trip recovery experiments

A trip recovery experiment was designed to gain more insight into the movement strategies used by individuals for trip recovery and to obtain input data for the trip recovery model. Data were collected using a group of younger and a group of older females.

3.1. Pilot testing

Initially a pilot experiment was performed to provide a good understanding of the external forces acting during the tripping protocol, and to ensure the forces acting on the subject were comparable to those expected in normal, daily living activities. The results of this experiment were used to optimise the safety and comfort of the subject. A secondary aim of this experiment was to optimise the distance from the tripping device to the force plate. The forces and impact acting on the subject's torso and feet were estimated when walking, jogging, successfully recovering from a trip, and when failing to recover from a trip. The acceleration of the head was estimated to assess the risk of a whiplash injury.

3.1.1. Methods and materials

Gait was obstructed with a tripping device, which consisted of two flat, rigid metal bars, (attached to a force plate [Kistler, 9287BA]) oriented along the direction of the walkway. Between these bars another flat, hinged metal plate was attached perpendicular to the walkway. This plate could be manually rotated upward (about 5 cm high) with a rope to obstruct the foot and induce a trip. The tripping device was directly attached to a force plate, which enabled the direct measurement of the forces acting on the device in the horizontal direction. The force plate was used to measure the horizontal force of the trip perturbation, and the ground reaction forces and duration of the initial recovery step.

A safety harness with full trunk support was used to prevent the subject from falling. The harness was attached with elastic ropes to an overhead I-beam and trolley system. Following an unsuccessful trip recovery, the elastic ropes elongated and slowed the rate of descent, reducing the impact on the subject. During normal walking and tripping, the harness supported the subject minimally and also influenced gait minimally (13% and 19% of body weight was supported by the harness for walking and tripping respectively). A load cell (Kistler 9331B) was placed in series between the harness and the trolley to measure the patterns of the total load placed on the harness system. A video camera was positioned perpendicular to the walkway to record the movements of the subject during trip recovery.

A younger female subject was used for the pilot experiment. CODA motion analysis IRED markers (Charnwood Dynamics Ltd.) were attached to the skin on the following landmarks (on the left side of the body): head (two markers: front and back), C7, head of the humerus, medial epicondyle of the elbow, greater trochanter, joint space of the knee joint, lateral maleolus and head of the fifth metatarsal. The subject was asked to perform the following trials:

Normal walking: The subject walked normally over the walkway and force plate. No CODA data were acquired.

Jogging: The subject jogged over the walkway and force plate. No CODA data were acquired.

Successful trip recovery: The subject walked over the walkway; gait was obstructed causing the subject to trip. The subject recovered with one step on the force plate, and made several more steps if necessary, resembling the trips that will be induced in the trip recovery experiment.

Failed trip recovery: The subject walked over the walkway, tripped, made no attempt to recover and was caught by the harness. This measurement simulated when the subject was unable to recover successfully.

Unweighting: The subject stood in the middle of the walkway and suddenly put all her body weight on the harness by lifting the feet of the ground.

The force plate, load cell and video data were acquired in six trials for each of the situations, CODA data were acquired in the trip trials only (200 Hz). Head accelerations during a normal walk were calculated from CODA data of the tripping trials before a trip occurred.

3.1.2. Results

Comparison with normal daily activities

Figure 3.1 shows a comparison of the vertical ground reaction force patterns of foot contact during a recovery step after a trip, a walking step, and a jogging step. The graph shows that the peak vertical forces acting on the contact foot while recovering from a trip were of similar magnitude to jogging. The initial slope of the force production, the loading rate, was higher than during walking, but about the same as when jogging. The time over which the force was produced was comparable to that during walking.

This means the impact on the recovery leg during trip recovery is comparable to that of normal, daily activities such as walking and jogging.

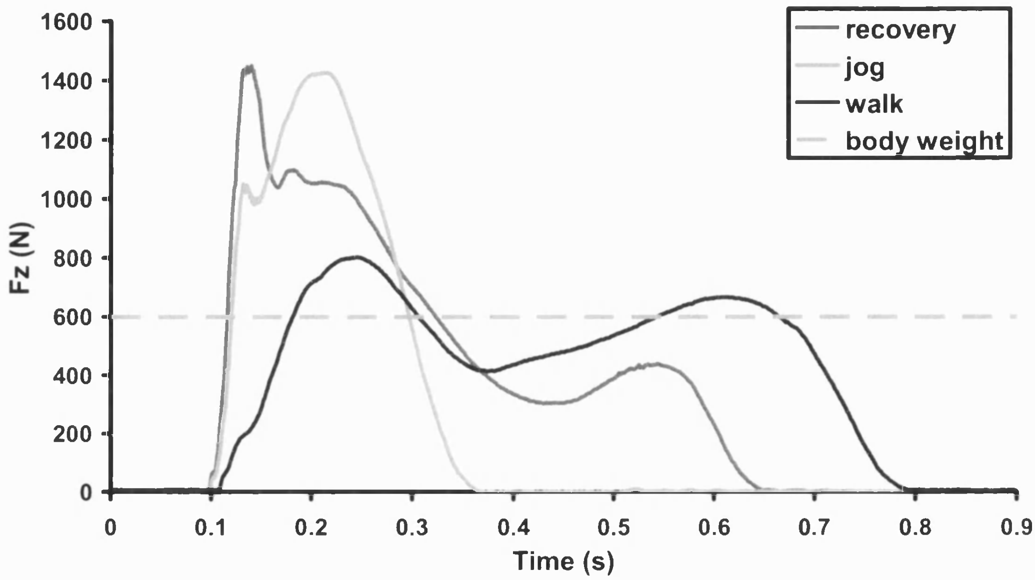


Figure 3.1 Vertical ground reaction force patterns during a recovery step, a walking step, and a jogging step.

Forces on the harness

Figure 3.2 shows tension forces acting in series with the harness during an unweighting trial, a successful recovery, and a failed recovery. The graph shows the forces from the harness acting on the body during successful recovery were minimal. The forces during a failed recovery were smaller than during a sudden unweighting trial, probably because ground contact remains during the attempted recovery. The force during failed recovery fluctuated around the value of one body weight. Initially the loading rate was similar, and later it was smaller than during a sudden unweighting trial.

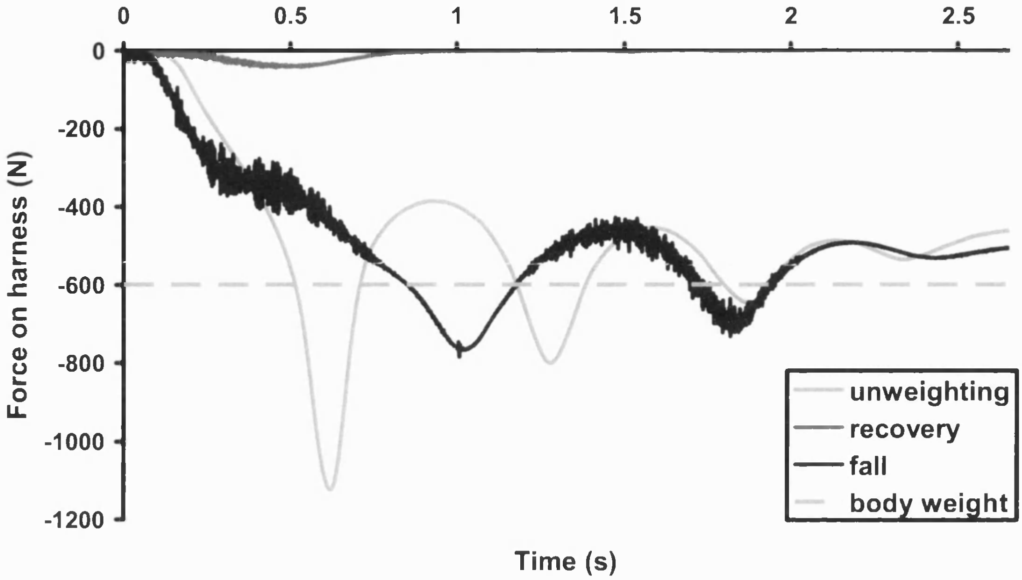


Figure 3.2 Force acting on the harness during an unweighting trial, a successful recovery and during a failed recovery.

Neck accelerations

Table 3.1 shows the mean peak neck angular acceleration in the sagittal plane both for a normal walk and a trip. The peak neck acceleration for a trip is about 2-3 times larger than during normal walking. However, values from literature (Keshner, 2003) show neck angular accelerations of about the same magnitude during sled decelerations without the occurrence of any injury. Head accelerations in situations in which whiplash injuries occur have been found to be substantially larger, and are over 234053 $°/s^2$ (Ivarsson et al., 2003). The risk of whiplash during the trip experiments was therefore concluded to be minimal.

Table 3.1 Mean peak neck angular acceleration in the sagittal plane for a normal walk and for a trip with standard deviations. It shows the mean over 7 trials in each situation. The last two columns contain values found by Keshner (2003) for sled decelerations.

	Walk		Trip		Keshner (2003)	
	Anterior	Posterior	Anterior	Posterior	Anterior	Posterior
Peak neck acceleration (°/s^2)	2278 ±407	2728 ±546	6875 ±1550	4908 ±1294	4800	4800

3.1.3. Conclusions

It can be concluded that the external forces acting on the body and resulting segment accelerations during the trip recovery experiment were similar to those experienced during normal daily activities and those shown by previous research to not cause injury.

The experimental set-up was modified slightly to that of the pilot testing and was subjected to a risk assessment by the department's technical officer and health and safety liaison (Appendix A1). The final tripping protocol was subjected to independent external review by Professor Julie Steele of the University of Wollongong, Australia (Appendix A1).

3.2. Inertia measurements

Inertia parameters are necessary as input for any model using equations of motion and were required for each subject. Subject-specific methods have the potential to provide more accurate values for body segment inertia parameters than regression methods based on cadaver data. The trip recovery experiment (section 3.3) had both older and younger subjects, and since body composition and therefore inertia parameters change with age, an accurate method was required that could be used for both younger and older women, which was not too invasive and time consuming for older people.

Two methods from the literature were compared prior to the experiments, one developed by Pavol et al. (2002) and the other by Yeadon (1990). The Yeadon method was developed for use with young gymnasts, while the Pavol method was developed specifically for use with older adults. Pavol et al. (2002) combined several existing methods to estimate segment masses and centre of mass positions, making use of a combination of body mass and segment length proportions, linear and non-linear regression equations and a mathematical model. The method requires only 32 measurements and was partly based on the modelling method of Yeadon (1990). The Pavol method is able to predict the body segment mass and centre of mass location of older adults accurately (Pavol et al., 2002). To determine if the Pavol method was also accurate for younger subjects, inertia parameters for two younger female subjects were calculated using the method of Yeadon (1990) and the method of Pavol et al. (2002) and compared.

The method developed by Pavol only described the calculation of segment mass, volume and centre of mass and not the calculation of the moments of inertia. A Matlab (Mathworks, Release 14) program was written, based on Yeadon's equations to extend Pavol's method to calculate the segment principal moments of inertia. Additional anthropometric measurements had to be taken to those described by Pavol (acromion width and depth, hip width and depth, knee perimeter, foot arch perimeter, foot nails perimeter, upper arm length and perimeter, elbow circumference, forearm length and perimeter, wrist perimeter, thumb perimeter, hand nails perimeter and hand length), bringing the total number of measurements to 48.

The masses calculated with both methods were compared with the measured total body masses and the errors were calculated. The Pavol method underestimated the total body mass for both subjects, while the Yeadon method overestimated it for both subjects. The Pavol method had a mean error of 1.1%, while the Yeadon method had a mean error of 5.2%.

The Pavol method was the most accurate in predicting the total body mass. There was substantial variation in segment mass and segment percentage mass values from the literature; Table 3.2 shows how values from the literature compare with values calculated with the Pavol and the Yeadon method. The marked values in the table agreed best with the mean of the values found in literature.

Given that it is difficult to establish a 'gold-standard' criterion against which to evaluate different inertia calculation methods, it was concluded that the Pavol method would be most appropriate due to better estimation of total body mass, its previous use with older adults, and its ease of application.

Table 3.2 Comparison of the percentage mass of the body segments found in the literature and measured with the Pavol and the Yeadon method. Column A are values by Braune and Fischer (1889) B by Dempster (1955), C by Clauser et al. (1969), D male data by Matsui (1958), E female data by Matsui (1958) all cited by Hay (1973). The values for both subjects are shown in columns I and II, those with a grey background are those that agreed best with the mean value of those found in literature.

								Pavol		Yeadon	
	A	B	C	D	E	mean	sd	I	II	I	II
Head	7.0	7.9	7.3	6.2	7.8	7.2	0.8	7.3	8.8	5.0	7.0
Trunk	46.1	46.9	50.7	48.7	47.9	48.1	1.6	48.9	41.5	44.6	44.3
Arm	3.4	2.6	2.6	2.6	2.7	2.8	0.0	2.9	2.9	3.0	2.8
Forearm	2.1	1.6	1.6	1.3	1.5	1.6	0.1	1.4	1.4	1.4	1.4
Hand	0.8	0.6	0.7	0.6	0.9	0.7	0.1	0.5	0.5	0.5	0.4
Thigh	10.7	9.7	10.3	11.2	10.0	10.4	0.7	10.9	10.9	13.9	13.3
Leg	4.8	4.5	4.3	5.4	5.4	4.9	0.6	5.0	8.1	5.5	5.4
Foot	1.7	1.4	1.5	1.5	1.9	1.6	0.2	1.2	1.0	0.9	1.1

3.3. Experimental data collection

3.3.1. Introduction

Experimental data were collected to provide data to answer the specific research questions posed in chapter 1 and later addressed in chapter 6, and to provide input data for the trip recovery simulation model. The research questions are restated briefly for reference:

1. What is the contribution of the recovery limb in successful trip recovery in both younger and older adults?
2. How do muscle sequencing and coactivation influence successful trip recovery in both younger and older adults?

3. What is the contribution of arm movement to successful trip recovery in both younger and older adults?
4. What is the difference in joint range of motion of the lower limb between younger and older adults, and how does this range of motion influence trip recovery?
5. How does the recovery step length vary in relation to trip recovery strategies in both younger and older adults?

3.3.2. Methods and materials

Subjects

There were two subject groups, one group of 20 to 35 year olds (n=8) and another group of 65 to 75 year old (n=7) females. It was decided to use female subjects only to eliminate potential gender effects, and because women on average fall more often than men (Pavol et al., 1999b). The first set of experiments was with the younger adults. Ethical approval was granted by the Bath NHS Local Research Ethics Committee, initially for younger subjects only (04/Q2001/169) and later for the older subjects (05/Q2001/214).

Body mass and height were similar for both groups (Table 3.3). The younger subjects were recruited through university and personal contact. The older subjects were recruited by contacting groups in the local community with older members (such as Lifeskills in Bristol, Ramblers Club Bath, and RICE Bath). Potential subjects were sent an information sheet about the experiment (Appendix A2 and Appendix A4). After showing initial interest in taking part in the study, an information letter (Appendix A3 and Appendix A5) was sent to the subject's GP to ask whether they knew of any reasons why this person should not take part in the experiment. Inclusion criteria were for the subjects to be recreationally active, community-dwelling and apparently healthy. BMI (body mass index) had to be below 28, and subjects must not have been on medication that causes dizziness at the time of the study, have any history of repetitive falling or show any fear of falling. Good health and absence of fear of falling was demonstrated by answering a health questionnaire (Par-Q (Thomas et al., 1992)) and the SAFFE fear-of-falling questionnaire (criterion score <0.75) (Lachman et al., 1998) (Appendix A8). The Par-Q questionnaire checked for the absence of neurological (stroke), musculoskeletal (osteoporosis, osteoarthritis), cardiovascular (high blood pressure, heart condition), pulmonary and cognitive disorders.

Table 3.3 Age, mass and height with standard deviations for both subject groups.

	Age (years)	Mass (kg)	Height (m)
Younger subjects (n=8)	26.1 ± 3.5	63.2 ± 8.4	1.67 ± 0.04
Older subjects (n=7)	70.0 ± 2.5	64.2 ± 4.8	1.66 ± 0.06

Inertia measurements

Anthropometric measurements were taken to allow determination of body segment inertia parameters using the modified procedure of Pavol (2002) as described in section 3.2. Body mass was measured using beam scales. Lengths, depths, width and perimeters of the body segments were measured with tape-measures and calipers.

Tripping device

In the final design of the tripping device, step-length data from the pilot experiment were used to determine the spacing between the obstacles and the force plate. The tripping device consisted of two metal rails (5 mm height) bolted to the force plate (in the direction of the walkway). Metal plates were attached between these rails. Eight metal plates of 0.1 m high were attached on these rails with hinges (Figure 3.3), four to obstruct the right foot and four to obstruct the left foot. The plates were positioned 25 cm apart in the direction of the walkway. These plates were spring loaded and held down by solenoids.

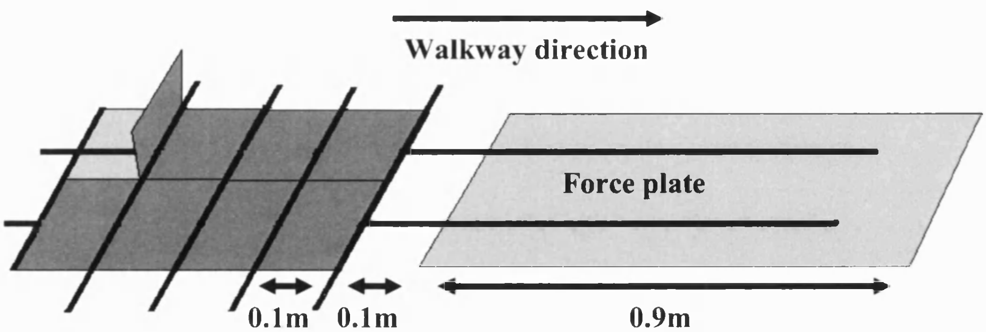


Figure 3.3 Schematic representation of the tripping device.

The plates could be released to obstruct the gait by activating the solenoids, after which the plate flipped up. The tripping device was directly attached to the force plate (Kistler, 9287BA); this enabled the direct measurement of the horizontal impact force acting on the subject's foot to induce the trip. There was friction between the tripping device and the floor creating a difference between the actual force on the foot and the force measured by the force plate. A calibration procedure measured the percentage of force lost by friction. For this calibration a load cell (Kistler 9331B) was attached to one of the obstacles of the tripping device and pulled with a constant force, to resemble the force acting on the obstacle when a trip is induced. The force measured by the load cell was compared to the force measured by the force plate. This was repeated five times and an average correction factor was calculated for the force lost to friction on the force plate measurements (32.5 ± 9.6 %). During the experiment the ground reaction force of the recovery step and the horizontal force on the tripping device were measured with the force plate sampling at 1000 Hz.

Safety measures

The safety harness preventing the subjects from falling in case they could not recover their balance was attached to an overhead I-beam and trolley system (Figure 3.4). This harness was specifically designed for gait rehabilitation devices and was a full torso harness. The rail to which the harness was attached was part of a scaffold structure built over the walkway. The harness was connected to a trolley on the overhead rail with karabiners and elastic ropes. This elasticity would decelerate an unsuccessful trip recovery and reduce the impact on the subject. Force in series with the harness was measured with a load cell (Kistler 9331B, 1000 Hz) to ensure the subject's body was not supported by the harness during apparently successful recoveries.

Subjects wore sport shoes with holes in them for IRED marker placement directly on the fifth metatarsal head and heel. A sports wrap-around ankle support, which allowed the ankle to move freely, was provided in line with recommendations of the ethics committee. Toe protectors were worn inside the shoes to protect the toes during impact with the tripping device.

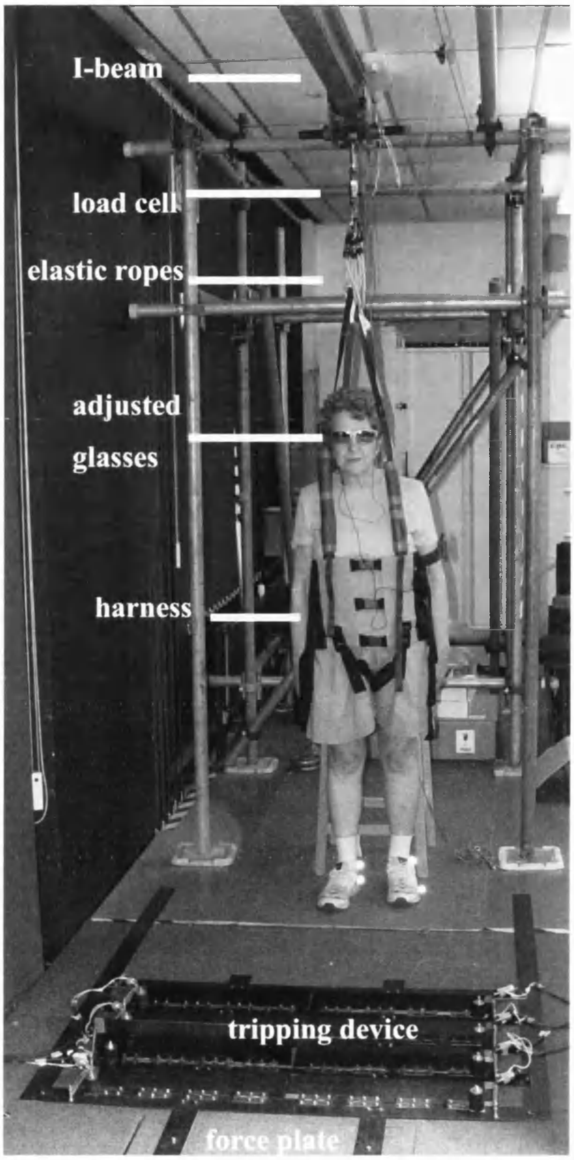


Figure 3.4 Photograph of the setup of the trip-experiment.

Kinematics

Whole-body kinematic data were collected with a CODA CX1 system (Charnwood Dynamics Ltd.) sampling at 200 Hz and positioned to the left of the subject. Three-dimensional information was collected using a custom-made triad set-up in combination with the Codamotion segmental gait analysis marker set. The triad setup consisted of rigid fibre board triangles with three CODA markers attached. Triads were used to define the position and orientation of rigid bodies where joint centres could not be viewed. In static trials CODA markers were temporarily placed either side of the joint, with the estimated joint centre in the middle, in order to calculate the relative position of the joint centres to the triad markers. This allowed the definition of virtual markers in dynamic trials. Marker triads were placed on the right foot to define position of the toes, on the right shank to define position of the ankle and knee joint centres, on the left and right forearm, to define joint centres of the wrists and elbows and on the chest to define the joint centres of both shoulders.

The Codamotion segmental gait analysis extensions formed a frame around the pelvis with markers on the side near the PSIS and as far forward as possible (ASIS) and an extension backwards placed near the sacrum (sacral wand). Another extension was wrapped around the left shank and head markers on a wand at the side of the leg (posterior tibia and anterior tibia); the same was done at the upper leg (posterior femur and anterior femur). Single markers were placed on the lateral aspect of the medio-lateral axis of the left knee, the lateral maleolus, the lateral side of the heel and the head of the fifth metatarsal. A single marker was placed on top of the head using a Velcro band around the head and a second Velcro band over the head. The joint centres of the left ankle and knee and both hips were defined with the Codamotion segmental gait analysis software, which required measurements of joint widths and segment lengths. A schematic picture of the marker placement is shown in Figure 3.5.

EMG

Bipolar EMG electrodes were used to measure muscle activity. Electrodes were placed on the following muscles: rectus femoris (RF), tibialis anterior (TA), gluteus maximus (GM), vastus lateralis (VL), semimembranosis (SM) and medial gastrocnemius (GA). The position of the muscle bellies were located via palpation and visualisation, and marked on the legs. The skin was cleaned with alcohol wipes before the electrodes were attached. The EMG signals were recorded at 1000 Hz with the Noraxon Telemyo wireless transmitter system. The wires of the electrodes and of the CODA system were taped to the body with an elastic bandage that was adhesive to its own surface only (Coban, 3M).

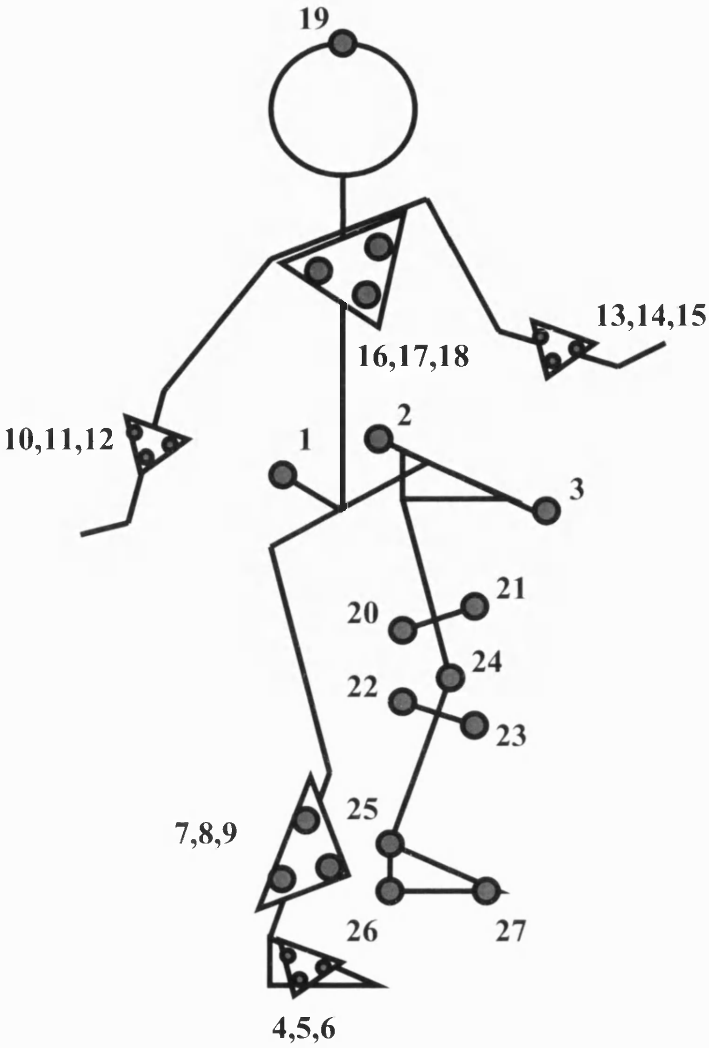


Figure 3.5 Placement of the CODA markers for dynamic trials; 1 on the sacral wand, 2 on the PSIS wand, 3 on the ASIS wand, 4, 5 and 6 on the right foot triad, 7, 8 and 9 on the triad on the right lower leg, 10, 11, 12 on the right forearm triad, 13, 14 and 15 on the left forearm triad, 16, 17 and 18 on the chest triad, 19 on top of the head, 20 on the posterior femur wand, 21 on the anterior femur wand, 22 on the posterior tibia wand, 23 on the anterior tibia wand, 24 on the lateral aspect of the medio-lateral axis of the left knee, 25 on the lateral maleolus, 26 on the lateral side of the heel and 27 on the fifth metatarsal of the left foot.

High speed video

To be able to calculate the spring and damper parameters to be used for foot contact in the trip recovery model, accurate data on the deformation of the heel and the forefoot during ground contact were required. In selected trials foot contact was recorded with a high speed video camera (Redlake MotionPro HS-1) with a sampling rate of 1000 Hz and a resolution of 768 by 604 pixels. The camera was placed perpendicular to the walkway with the centre of the force plate in the middle of the field of view. The field of view was approximately 1 m wide, which resulted in an accuracy of approximately 1.0 mm (0.1% of the field of view). For the calibration, four markers were distributed around a planar rigid frame and their locations were measured accurately with a tape measure. Two spotlights provided the extra lighting required for high speed filming. The CODA system works with infrared signals to get the position of the markers and this signal was disturbed by the spotlights used for the high speed camera. For this reason the high speed camera trials were performed separately from the trials with CODA and EMG. All trials were filmed with a digital video camera (Sony DCR-TRV-900E) for qualitative assessment of the trials and selection of trials for future analysis.

Experimental protocol

Prior to the experiment, the experimental procedures were explained and an informed consent form (Appendix A6) was signed by both the subject and the investigator. The subject was reminded that she was free to withdraw from the study at any time, without giving any reason. Questionnaires were completed and checked to ensure the subject met the inclusion criteria. The questionnaires had been sent to the subject to familiarise them prior to the experiment. Anthropometric measurements were taken in accordance with the modified Pavol (2002) method and the subject was prepared with the CODA markers and EMG electrodes. The subject was strapped in the harness that was adjusted for a tight but comfortable fit. The subject familiarised herself with the harness until she felt safe putting all her weight in the harness and swinging around. Trips were not practiced to prevent the subject from getting used to the trip stimulus. The subject was given prescription-free glasses with the lower half obscured to prevent her from seeing when the tripping device was activated. A portable music player with inner ear headphones prevented the subject hearing the plates of the tripping device flip up when activated.

CODA, force plate, load cell and EMG data were collected when the subject walked over the walkway at a self-selected pace. In random trials, a trip was induced by activating the tripping device. Fifty percent of the trials were trip trials. The first 50 trials were completed with EMG electrodes on the left leg, the second set of 50 trials with electrodes on the right leg, with trips being induced on both legs in each set. After this the EMG electrodes and CODA markers were removed from the subject's body and 15 trials were performed with high speed video recording (1000 Hz), the majority of these trials were trip trials. These trials were only used to obtain data for the ground-contact elements of the simulation model.

During the first sets of trials EMG, CODA, and force plate signals were synchronously collected on a PC using the Codamotion software (Version 6.69) and load cell data were collected on another PC using Kistler software (Bioware, version 3.2.6.104). The load cell data collection was triggered and synchronised by a hardware trigger produced by the CODA system at the start of the data collection.

During the high speed camera trials, force plate and load cell data were synchronously collected on the same PC using the Kistler software and high speed video data were collected on a laptop running acquisition software (Redlake MotionPro Central, V1.12). High speed video data were only collected around foot contact. The high speed video recording was triggered by an LED light trigger box (Wee Beastie, UK), which also sent out a trigger to the PC recording the force and load cell data. The trigger was manually induced with a remote control. It was recorded as a voltage signal and produced a time stamp on a spare analogue channel.

3.3.3. Protocol changes for the older subjects

In line with recommendations from the Ethics Committee, some changes were made to the experimental protocol for use with the older subjects. An extra inclusion criterion was added; the main risk factors for osteoporosis had to be absent. This was checked with the form for referral for a DEXA (bone mineral density) scan used in the Royal National Hospital for Rheumatic Diseases (RNHRD bone densitometry direct referral-dexa scan, Appendix A9).

It was expected that the older adults would have slower responses during the trip recovery trials than the younger adults. These slower responses can be caused by slower detection of the trip stimulus, reaction time to the trip stimulus and by a slower movement velocity. To distinguish between these three underlying causes, foot sensation and response time were measured in the older adults. With foot sensation measurements the presence of sensory degradation could be indicated, which can cause a slowed detection of a trip stimulus. The foot sensation measurements and response test times were correlated with trip recovery experiment outcome measures to investigate their effect on trip recovery. Foot sensation and response time were measured in the older adults only, as it was assumed that sensory degradation and slowed response were absent in the younger adults.

Foot sensation was tested using Semmes-Weinstein monofilaments (Rolyan monofilaments). Filaments with different grading were pressed into a c-shape on different parts of the foot (hallux, first and fifth metatarsal, mid-foot and heel) and the subject was asked if, and where, she could feel this. Sensation testing was completed with the subjects lying prone with knees flexed to approximately 90 degrees to prevent them from seeing what was happening. The monofilaments could distinguish between normal sensation (equivalent to 0.41 grams), protective sensation (equivalent to 2.05 grams), and loss of protective sensation (equivalent to 29.00 grams).

Response time was tested by a single step test after a visual input, as a step response is close to the type of response necessary to recover from a trip. The visual trigger in the response test differs however from that when a trip occurs, which is a sensory trigger. A line was placed in front of the force plate and a cross was placed on the centre of the force plate. LED lights were placed behind the force plate, facing the subject when she stood behind the line. The subject was asked to stand behind the line and step with the preferred foot on the cross. A trigger was sent out when the LED lights were turned on and the signal was collected by a PC using Kistler software. Force on the force plate was collected on the same PC. Response time was defined by the time between the lights coming on and the first contact with the force plate.

To ensure the experiment was not too strenuous for the older subject group, it was spread over two days and the number of trials with CODA and EMG was reduced from 100 to 60. To obtain enough useful trip trials the percentage of trips of these trials was increased from 50% to 67%. On the first day the experiment was explained and the informed consent forms and questionnaires were completed. After ensuring the subject met the inclusion criteria, anthropometric measurements were taken and the foot sensation and response tests were performed. Ten to fifteen trials were collected with the high speed video, the number of trials depended on the attempts required to get data for all recovery strategies. On the second day the 60 trials with CODA, EMG and load cell data were collected.

3.4. Data processing and data analysis

A Matlab routine was written for the main data processing. Each trip recovery trial was different in timing, magnitude of the perturbation and recovery success. This made it difficult to calculate mean values for all the different trials. For this reason typical trials were chosen for each subject and for each strategy and recovery leg. The criteria for choosing these trials were to choose those that were representative for that strategy, that had the fullest set of data, and that had single foot contact on the force plate with the recovery foot.

A Matlab subroutine was written to calculate the inertial parameters using the method developed by Pavol et al. (2002) using the subject-specific anthropometric data.

Calculating joint centre positions in CODA

Joint centre positions of the left leg were calculated using the CODA segmental gait analysis software and specific anthropometric data. The joint centres of the remaining joints were calculated using the triad marker and joint centre positions from the static trials. Virtual marker positions, representing the joint centres, were calculated as weighted averages with a constant offset from the 3-D positions of the triad markers. All raw joint centre coordinates were exported to text files and read in by the Matlab analysis routine.

Force data

The coordinate system of the force traces was as such that a positive Fz was in an upward direction, a positive F_y in an forward direction and a positive F_x was directed to the right. Net forces and the mean and standard deviation of the force traces in the first second of the trial, before the trip, were calculated. To define when a change in force occurred, a reference value (ref) of the mean plus four standard deviations of the first second of the force data was calculated (for F_x mean ref = 1.71 N, for F_y mean ref = 2.55 N, for F_z mean ref = 7.27 N). Force plate contact was defined by using F_z (force in the vertical direction). This was done by differentiating F_z and finding when this value was almost zero, resulting in the peak F_z (F_{zpeak}). From the F_{zpeak} the contact with the force plate was found by going backwards in time and finding when F_z dropped below ref. Loss of contact with the force plate was found in a similar way, only looking forward in time. The force signal contained additional signal after contact with the tripping device due to vibrations, until the foot made contact with the force plate. F_z was used to define contact as the noise was minimal in F_z. Force peaks of the other signals F_y (posterior-anterior horizontal force), F_x (medio-lateral horizontal force) and F_{lc} (force measured with the load cell) were found in the same way as F_{zpeak}, looking at the force signal during the contact phase only. F_y and F_{lc} were reversed at the start of the data analysis, which made all force in a forward or upward direction positive.

Each force signal had different peaks (Figure 3.6); F_x had a positive peak with ground contact (F_{xfc}) and a negative with push-off (F_{xpo}); F_y had negative peaks on contact with the tripping device (F_{ytc}) and ground contact (F_{yfc}) and a positive with push off (F_{ypo}); F_z had a positive peak with ground contact (F_{zfc}); F_{lc} had a positive (tension) peak just after the trip (F_{lctr}). Mean and peak loading rates up to the peak forces were calculated. The mean loading rate was calculated by dividing the difference between the peak force and ref by the time over which the peak force was created. The maximal loading rate was calculated by finding the maximal derivative of the force trace during the time period between contact and F_{zpeak}, or the minimal derivative for a negative peak. The impulse of each peak was calculated using the trapezium rule. This method was compared with some more complicated and computationally intensive rules (Bode's rule and Simpson's rule) with several functions with a known integral. All integration techniques gave similar results, so the trapezium rule was chosen because of its simplicity and its accuracy was comparable to the other methods.

During contact with the tripping device, friction between the tripping device and the walkway caused a difference between the actual force on the tripping device and the measured force on the force plate. Calibration measurements in pilot testing showed this difference was a mean of 32.5±9.6%. This percentage was added to F_y during contact with the tripping device.

The maximum percentage weight supported by the harness during trip recovery was calculated. A trip trial was considered to be an unsuccessful trip recovery when more than 30% of the body weight was supported by the harness. This percentage was based on Madigan and Lloyd (2005).

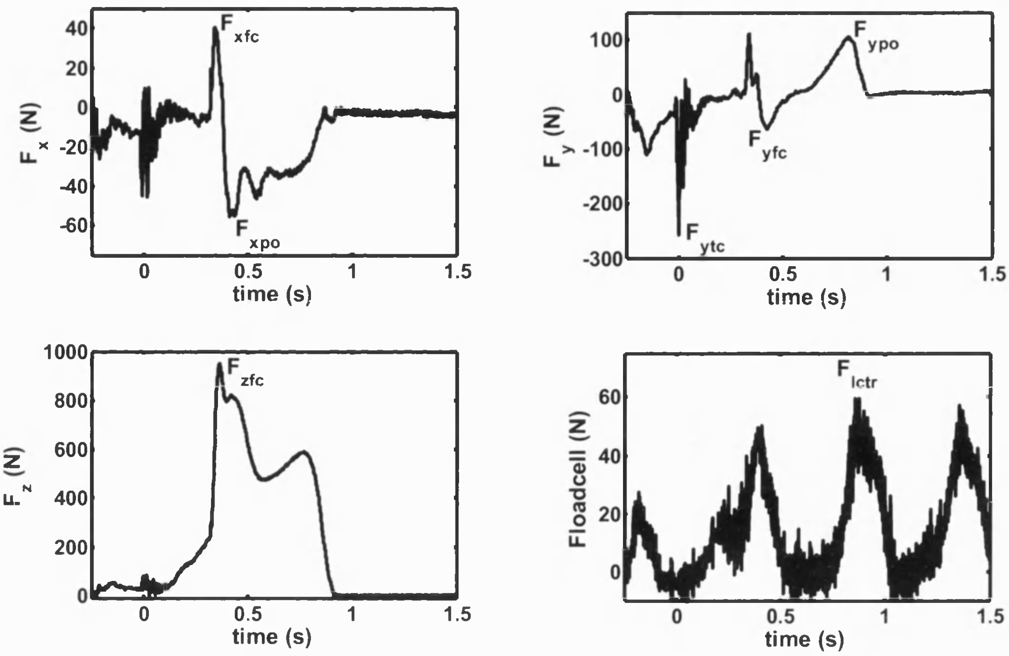


Figure 3.6 Force traces with peaks, F_{xfc}: positive peak in F_x with ground contact, F_{xpo}: negative in F_x with push-off, F_{ytc}: negative peak in F_y on contact with the tripping device, F_{yfc}: negative peak in F_y at ground contact, F_{ypo}: a positive in F_y with push off, F_{zfc}: positive peak in F_z with ground contact and F_{lctr}: positive (tension) peak in F_{lc} just after the trip.

Synchronising trials

The time a trip was induced varied in the trials. To be able to compare trials, all trials were synchronised by taking the time of contact with the tripping device (F_{ytc}) as zero. The period 0.25 s before to 1.50 s after the time of contact was processed for further data analysis. This allowed analysis of part of the walk preceding the trip and the complete trip recovery step.

Data smoothing

CODA marker positions were smoothed and interpolated with the Woltring's B-spline using the GCVSPL Matlab routine written by Reina (1998). This method used the generalised cross-validation and mean-squared prediction error criteria of Craven and Wahba (1979). A fifth order (or quintic) spline was selected for use. The smoothing procedure was non-iterative and a value for the smoothing parameter P had to be given.

After residual analysis investigating joint angles and moment estimates a P value of $5e^{-8}$ was chosen (equivalent to cut-off frequency for marker position data of 39 Hz).

An interactive Matlab routine was written that allowed points to be inserted into regions where the markers were obscured to avoid outlying interpolated values. The routine allowed insertion of ten points into the area where the markers were obscured and steered the interpolation in the right direction. It was visually checked whether points needed to be inserted by comparing the interpolated marker positions to those of a similar trial. The interpolation routine was compared with the cubic interpolation available in the Codamotion software by removing data from an existing CODA marker position and comparing the interpolated values (Figure 3.7); this was done for different marker positions and different trials. It can be seen that the Matlab routine gave better results as the expected marker position is included in this routine. This expected marker position is based on the position in similar trials.

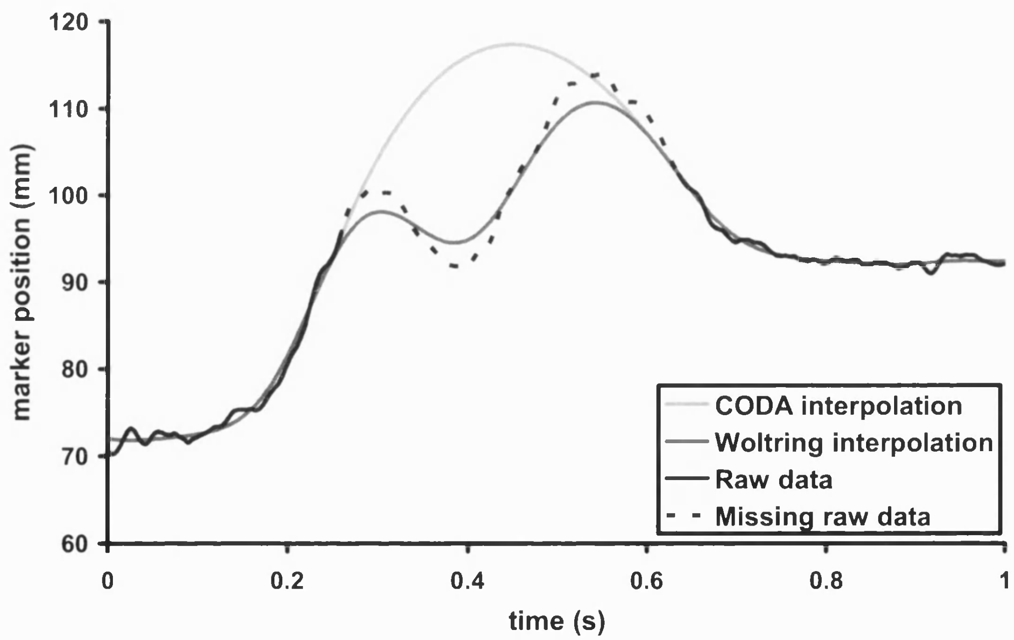


Figure 3.7 Comparison of marker interpolation with the interactive Matlab routine using the Woltring spline and marker interpolation with the cubic interpolation available in the Codamotion software.

No further smoothing was done in the data analysis routine on values derived from the CODA marker positions. First and second derivatives were calculated using the Woltring routine without any smoothing.

The smoothed coordinates were used to calculate the CM trajectories of the 17 body segments and of the whole body. Y-Z projection joint angle trajectories of the ankle, knee and hip were calculated from segment angles and defined so extension was in the positive direction and flexion in the negative direction ensuring continuity.

Joint moments, work and stiffness

Joint moments of the ankle, knee and hip were calculated using standard inverse dynamics techniques (Appendix C). Forces were resampled to 200 Hz. Joint moments could only be calculated for trials with single foot contact. Trials with the recovery foot only on the force plate were selected by looking at the videos of the trials and at the force traces. Time of foot contact on the force plate was defined by when F_z was above the reference value for F_z.

Angular joint power during contact with the force plate was calculated by multiplying the joint moments with their angular velocities. Work was calculated as the integral of the angular power, and cumulative work was calculated to give the total work done during a recovery step.

Torsional stiffness of the ankle, knee and hip (k_{ankle}, k_{knee}, k_{hip}) were calculated as the ratio of the mean rate of change in joint moment to joint angular displacement from ground contact until maximum joint flexion (Farley & Morgenroth, 1999), or until time of maximum joint torque if this occurred first.

External moment and angular momentum

The external moments of the body during the perturbation (contact with tripping device) and during trip recovery (contact with force plate) were calculated as by Pijnappels et al. (2004) (Equation 3.1).

Equation 3.1 $M_{ext} = \vec{F}_{gr} * \vec{d}_{gr} + \vec{F}_c * \vec{d}_c$

Where F_{gr} is the vertical ground reaction force at the CoP, F_c is the horizontal contact force and d_{gr} and d_c are the vectors from the body CM to the point of application of the force vectors. The time integral of the external moment is equal to the change in angular momentum.

The angular momentum about the centre of mass (CM) was calculated by the segmental method (Appendix B). In this method the angular momentum was calculated for each segment and summed to get the total body angular momentum. To calculate the angular momentum with the segmental method the angular velocities of the body segments were required and these were calculated by differentiating the segment angles.

To obtain an indication of how much the forward angular momentum induced by a trip was reduced during recovery the variable recovery amount (RA) was introduced. This variable was defined as the difference between the maximum angular momentum occurring between the start of the trip and foot contact with the force plate and the minimum angular momentum during foot contact with the force plate

Recovery step length

Medio-lateral, anterior-posterior and overall recovery step length were calculated using the ankle coordinates of the obstructed foot at contact with the tripping device and at contact of the recovery leg with the force plate.

EMG

The raw EMG signal was full-wave rectified. The EMG system had automatically applied an anti-alias filter and a high pass filter of 10 Hz to the signal.

The rectified EMG signals were normalised to the subject's mean EMG amplitude during a full stride of all walking trials. Some key events were found to identify a full stride; mid-stance and mid-swing of the same limb. Start of mid-stance was found by identifying when horizontal acceleration of the CM of the foot was zero preceded by a negative horizontal acceleration. Mid-swing was identified as the next time point when the horizontal acceleration was zero, preceded by a positive horizontal acceleration. The second mid-stance was the third time point when horizontal acceleration was zero, preceded by a negative horizontal acceleration. A full walking stride was identified by the first to the second mid-stance of the right leg occurring at least 400 ms after the start of the trial.

The EMG signal was used to calculate the timing and magnitude of muscle activation and muscle coactivation to consider muscle sequencing. An "on-off" signal was calculated which indicated when muscles were activated or not. To define when a muscle was activated a threshold value, above which the muscle was deemed active, was needed. This threshold value was defined as the mean EMG signal when the muscle was inactive multiplied by a factor which was subject to visual checking (this factor varied from 3 to 25). The mean signal during inactivity was taken as the mean of the first 300 ms of the trial. If the muscle was active at the start of the trial a period of inactivity was manually chosen. The factor with which the mean was multiplied to get the threshold was visually chosen by checking whether the "on-off" signal agreed with the processed EMG signal. The multiplication factor was muscle and subject dependent, as the background noise levels and EMG amplitudes vary due to differences in skin impedance and placement of the electrodes. The maximum peak of the EMG signal after the trip was induced was calculated together with the time after the trip at which it happened.

Coactivation of the tibialis anterior and gastrocnemius ($coact_{ankle}$) was found by calculating the integrated EMG of tibialis anterior for the period around impact when gastrocnemius was active and normalising this to this period of time (Hortobagyi & DeVita, 2000). The period around contact when gastrocnemius was activated was found by the "on-off" signal, if gastrocnemius was not active at impact the first activity period after impact was used to calculate coactivation. Coactivation of biceps femoris and rectus femoris ($coact_{knee}$) and of rectus femoris and biceps femoris ($coact_{hip}$) were calculated in a similar way.

Normalisation of variables

To annul the effects of body composition, and to be able to compare inter-individual results, some variables were normalised. Scaling methods in the literature are not consistent and there appears to be no accepted standard in this field. Distance variables and velocities such as base of support, centre of mass position, and walking velocity were scaled to lower limb length (LL), and arm movement was scaled to arm length (AL), making the distance variables dimensionless and velocities with a unit of s^{-1}. It was chosen to normalise the distance and velocity variables to LL or AL and not to body height (as is sometimes done in the literature) because they were expected to be more directly influenced by LL or AL respectively. Forces were scaled to body weight (BW), as this is a unit of force and makes the force dimensionless. Joint moments were scaled to body weight times lower limb length (BW*LL), which made them dimensionless. It was chosen to scale the moments to BW*LL as moment is the product of the moment arm and force;

force is correlated to BW and moment arm to LL. Angular momentum was scaled to body weight times lower limb length squared ($BW*LL^2$), which gives it the dimension sm^{-1}. It was chosen to normalise the angular momentum to $BW*LL^2$ to account for differences in segmental moment of inertia values.

Statistical analysis

Statistical analysis was performed in SPSS (version 14.0). Differences between younger and older, or trip and no trip trials, or elevating and lowering trials were investigated with an unpaired t-test for continuous variables and a chi-squared test for categorised variables. Associations between continuous variables were investigated using a Pearson test.

Chapter 4: Trip recovery simulation model

A torque driven linked-segment trip recovery model was developed to allow investigation into the contributions to successful trip recovery. To obtain more insight into trip recovery models and investigate what information could be derived from simple models two existing models were reproduced; an inverted pendulum model by van den Bogert et al. (2002), and an inverted pendulum model with springs by Hsiao and Robinovitch (1999).

4.1. Simple inverted pendulum model

4.1.1. Background

Van den Bogert et al. (2002) developed a single inverted pendulum model to investigate whether walking velocity or response time had more effect on preventing a fall. The human body was represented as a single rigid element with uniform density. The model assumed the body had no angular velocity prior to the trip and rotated about a fixed axis after the trip.

4.1.2. Simulation

The model was replicated in Matlab (version 6.5, release 13, Mathworks). Figure 4.1c shows the replicated tilt angle versus time graph from van den Bogert et al. (2002). Input constants were taken from the article: body height = 1.87 m, initial tilt angle = 9.3°, and a replicated initial horizontal velocity = 0.694 body heights/s. Inverted pendulum motion was simulated with different initial angular velocities (0-0.4 body heights/s), an initial tilt angle of 8° and varying response times (0-0.4 seconds). Simulation results are shown in Figure 4.1b and d and original results in Figure 4.1a and c.

4.1.3. Conclusions and discussion

The simulations successfully replicated the results of van den Bogert et al. (2002). These simulations showed that a simple model can be used to answer some questions about trip recovery. For example van den Bogert et al. (2002) concluded that response time was more important than walking velocity for successful recovery. The addition of a second segment with a spring, to represent the recovery leg, would bring the model closer to reality and enable it to investigate the effect of step length. The next paragraph describes such a model developed by Hsiao and Robinovitch (1999).

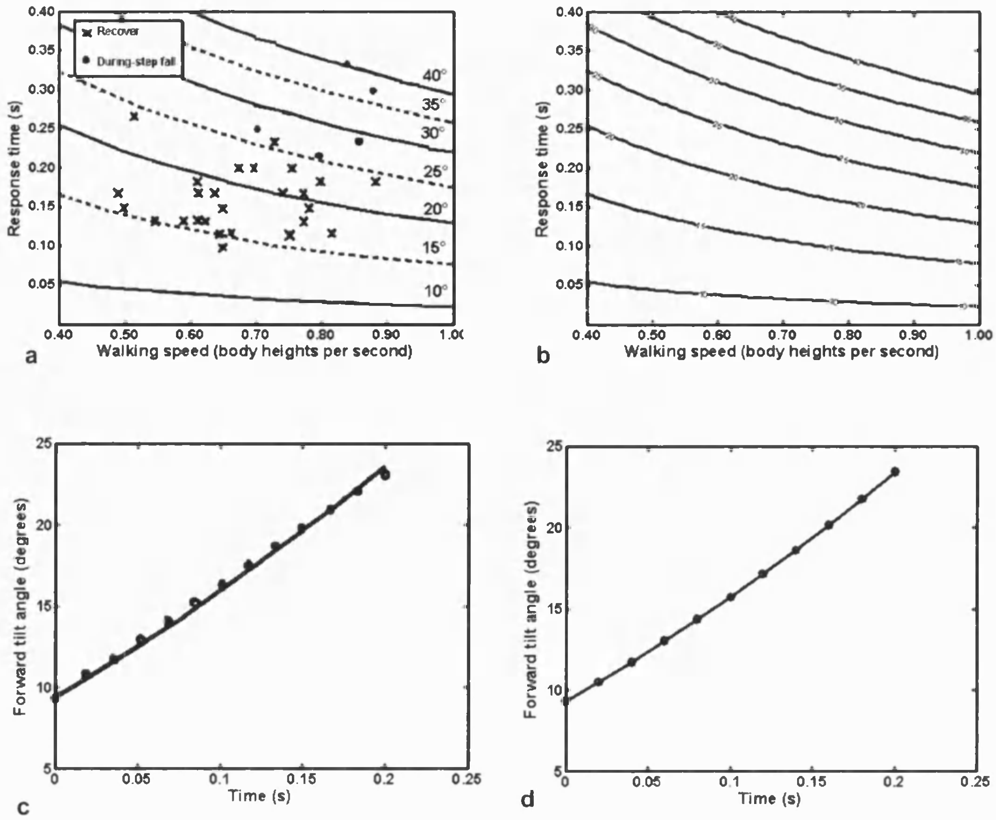


Figure 4.1 (a) and (b): Simulation of the response time versus the walking speed in body heights per second. Figure a is from Bogert et al. (2002), figure b is simulated with Matlab. (c) and (d): Simulation of the tilt angle in degrees versus time. Figure c is from Bogert et al. (2002), figure d is simulated with Matlab. The circles in figure c are experimental data; the circles in figure d are simulated data points.

4.2. Inverted pendulum model with spring

4.2.1. Background

A simple pendulum-spring model of balance recovery by stepping was developed by Hsiao and Robinovitch (1999) to assess how step length and step contact time influence the effort, leg contact force, and feasibility of balance recovery by stepping.

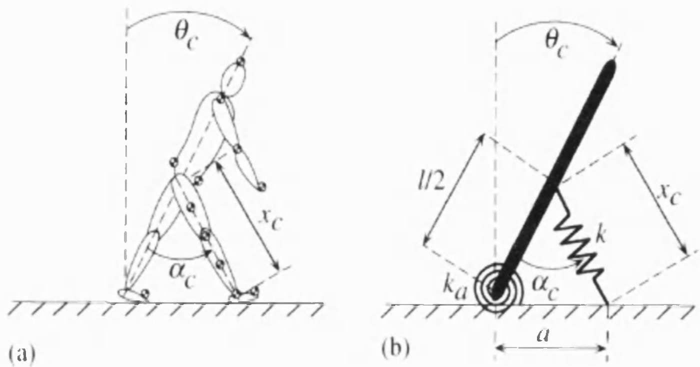


Figure 4.2 Experimental (a) and mathematical (b) models of balance recovery by stepping (Hsiao & Robinovitch, 1999).

The model simulated both pre-contact and contact phases, and consisted of three elements (see Figure 4.2): an inverted pendulum, representing the head, torso, upper extremities and pivot leg; a mass-less linear spring, representing the stepping leg; and a rotational spring, simulating the net effect on the body's downward rotation of lower extremity torques during the pre-contact phase of the step. The model assumed the pivot leg was fully extended at the instant the leg spring contacts the ground and that rotational spring constant was zero throughout the step contact phase. Horizontal distance between the feet was assumed constant throughout step contact.

4.2.2. Model

The model was solved as an initial value problem with initial conditions for the body angle and angular velocity. The governing equation of motion during pre-contact phase was based on pendulum motion and used the rotational spring constant. The moment of step contact was defined by the body angle reaching a threshold value and the time being equal to step contact time. The rotational spring constant became zero at this point and compression of the recovery leg spring was added to the equation of motion. The equations of motion were solved with a fourth-order Runga-Kutta integration routine with a time step of 4 ms. Successful balance recovery was signified by the occurrence of a negative body angular velocity whilst the body angle was below 90°. For these trials recovery effort was defined as the corresponding peak force and the energy absorbed in the leg spring. Body mass was 68 kg and body height 1.7 m. The rotational spring constant was 245 Nm/rad (based on experimental data), and the stiffness of the leg spring was 15 kN/m.

4.2.3. Simulation

The pendulum-spring model by Hsiao and Robinovitch (1999) was repeated through simulation using Matlab. The Matlab ode45 routine (fourth-order Runga-Kutta integration) was used to solve the equations of motion. An initial step size of 4 ms was used; ode45 automatically reduces step size when errors become too large.

After correcting the published equations, several examples of simulations provided in the original article were repeated using the Matlab model (Table 4.1). The values in the highlighted cells in Table 4.1 did not agree with the article. To obtain a better overview of when the simulations agreed with those of Hsiao and Robinovitch (1999), Figure 4.3a was replicated with the Matlab model (Figure 4.3b).

Figure 4.3b shows the simulations agreed with those of Hsiao and Robinovitch (1999) in some areas of the graph, and differed in others. The replicated values in the area where successful balance recovery was possible following the simulation, agreed well with values from the article. However, the area in which successful recovery was possible differed and the values outside the area differed from the article.

To replicate the solid lines in Figure 4.3a simulations, with a constant ratio of the body angle and stepping angle at contact, were performed (Figure 4.3c). This time the area of possible successful recovery found with the simulations agreed with the article (see Figure 4.3c). The values found within this area also agreed with the values from the article. There was only one deviating value for angle ratio and larger oscillations were found with the simulations. Choosing a larger stepping angle increment (e.g. 8° instead of 2°) reduced the oscillations, but also reduced the accuracy of approximation in the inter-lying points (see Figure 4.3d).

Table 4.1 Simulation of some examples from the article of Hsiao and Robinovitch (1999). With α_{ini}: initial body angle, t_{step}: step contact time, k_{rot}: rotational spring constant, r_{α}: angle ratio, α_{step}: stepping angle, α_{cont}: body angle at contact and F_{max}: maximal force. The values in the highlighted cells do not agree with the article.

	Input values					Hsiao and Robinovitch (1999)		Simulation	
	α_{ini}	t_{step}	k_{rot}	r_{α}	α_{step}	α_{cont}	F_{max}	α_{cont}	F_{max}
#2	14	460	0	1.71	51.3	30.0	1.9	28.7	1.60
	14	460	245	1.71	39.3	23.0	1.3	21.8	0.79
#3	8	690	0	1.73	53.6	31.0	-	30.7	-
	14	520	0	1.71	57.1	33.4	-	33.5	-
	23	450	0	1.59	72.8	45.8	-	45.3	-
	8	920	245	1.73	53.6	31.0	-	30.8	-
	14	690	245	1.71	57.1	33.4	-	33.2	-
	23	620	245	1.59	72.8	45.8	-	46.5	-
#6	14	400	245	-	39.0	-	1.1	-	0.61
	14	700	245	-	67.0	-	1.7	-	1.66
	14	700	245	-	39.0	-	2.0	-	1.89
	14	700	245	-	29.0	-	2.6	-	1.72

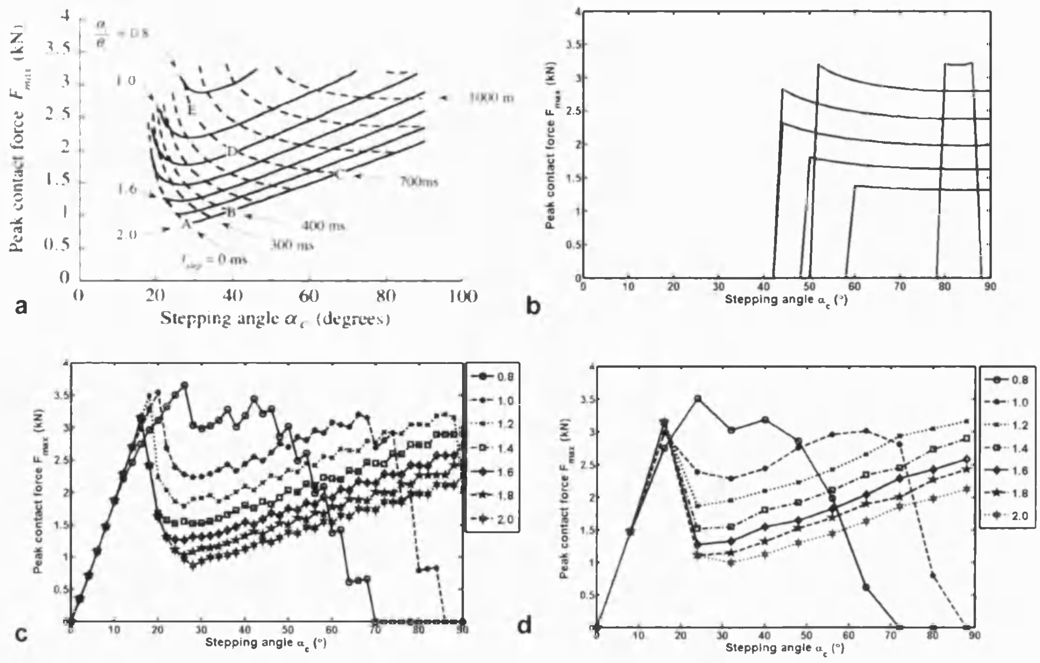


Figure 4.3 (a) Model predictions by Hsiao and Robinovitch (1999). Peak contact force in the stepping leg scales with the stepping angle and step contact time. Solid lines represent constant angle ratio and dashed lines represent constant step contact time. Any point within the mesh defines a combination of stepping angle, step contact time and angle ratio which allow for successful balance recovery. (b) Replication of the dashed lines (constant step contact time) in a. (c) Replication of the solid lines (constant angle ratio, step contact time increasing with steps of 2°) in a. (d) Replication of the solid lines (constant angle ratio, step contact time increasing with steps of 8°) in (a).

4.2.4. Conclusions and discussion

Not all results agreed with the original article (Hsiao & Robinovitch, 1999). Personal communication with the author (Robinovitch, 2004) failed to solve the source of these discrepancies.

The model was a simplification of reality and had limitations; it did not account for swing-phase dynamics; recovery effort was defined as the peak contact force and energy absorbed in the stepping leg during step contact, while energy was also expended during step initiation and swing; the model used a single point of contact to simulate impact of the foot with the ground.

Successful trip recovery is a combination of factors such as lower limb strength, response time, recovery step length, walking velocity, arm movement, lower limb kinematics and joint range of motion. To be able to investigate a combination of these factors a more complex trip recovery simulation model was developed. The resultant of leg strength was represented by joint moments of the ankle, knee and hip, response time by the activation timing of the joint moment generators and arm movement by the movement of the upper body CM, which was represented as a single mass moving relative to the pelvis. This simulation model progressed though a number of developmental stages before reaching its final design.

4.3. Bipedal knee-less walking model

A software package that assists in formulation of the equations of motion was used in the development of the trip recovery model. Several such packages are available; two of which were investigated in more detail, Autolev and Simmechanics. Autolev is based on Kane's method of formulating the equations of motion (Kane & Levinson, 1985) and is developed for engineering and mathematical analysis. Matlab is a general scientific computer language, and has a toolbox, Simmechanics, that assists in formulating the equations of motion. Matlab has the advantage that it contains toolboxes computationally strong in various areas. Autolev produces C or FORTRAN files that have to be modified to be integrated into a simulation routine.

A bipedal walking model was created in both Matlab and Autolev to compare both packages and gain experience in developing models. This model consisted of two single segment legs without mass. The feet remained in contact with the ground and friction with the ground was ignored. The model was angle-driven with prescribed angles of the hips and one foot. These angles were experimentally obtained in a pilot experiment. The Autolev model consisted of two different routines, each with one of the legs fixed, that were combined in a single C-routine. The Matlab model using the Simmechanics toolbox consisted of only one routine. The stance and the swing leg were defined by prescribing the trajectory of one foot and prescribing the angles of the hips and the ankles. Both models gave similar results for foot trajectories and torque output patterns. Simulations of the Autolev model were faster than with the Matlab model (approximately 40%). It was initially decided that further modelling would be performed using Matlab, because of its ease of use and the multiple toolboxes available.

4.4. Spring-damper model

In the trip recovery model foot-ground contact was modelled with multiple spring-damper systems at the foot-ground interface. To obtain more insight in modelling spring-damper systems with Simmechanics a simple spring-damper model was developed. The model was initially 1D, and was later extended to 2D and 3D. A custom spring-damper was created, which produced a reaction force when the distance between the two bodies connected to the spring-damper system was below a threshold. The force created by the spring-damper system was calculated with the equations for damped harmonic vibration, using spring and damper constants. The spring-damper system (e.g. heel pad) complied at ground contact, created an upward force, and finally moved upwards. The system stopped creating an upward force as soon as the bodies were a certain distance apart. The model was validated by comparing the simulation results with values analytically calculated with the equations, for varying spring and damper constants.

4.4.1. Simulation

A single contact step of a sprint was simulated to show the model was capable of simulating foot-ground contact. The body was represented by a mass of 70 kg positioned at a height of 0.9 m. The initial horizontal velocity of the body was 10.0 m/s, and the initial height of the foot was 0.24 m, based on sprinting data collected within our research group. A spring constant of 126.6 kN/m (Alcantara et al., 2002) and a damper constant of 100 Ns/m were used (Andrews & Dowling, 2000a). Simulation results are shown in Figure 4.4. The heel compliance is 55 mm, which is more than would be expected (1.9-6.3 mm (Aerts et al., 1995), 7.2-8.8 mm (Alcantara et al., 2002)). This may be because the spring-damper properties of the rest of the body were not accounted for in the model, and this caused the heel pad alone to dampen the body movement. The horizontal displacement was large, which might be caused by the spring-damper parameters, which were estimated from literature. The model used a linear spring which was active during the heel compliance phase, but the spring properties of a foot are often modelled as non-linear spring-dampers, such as by Wilson et al. (2006).

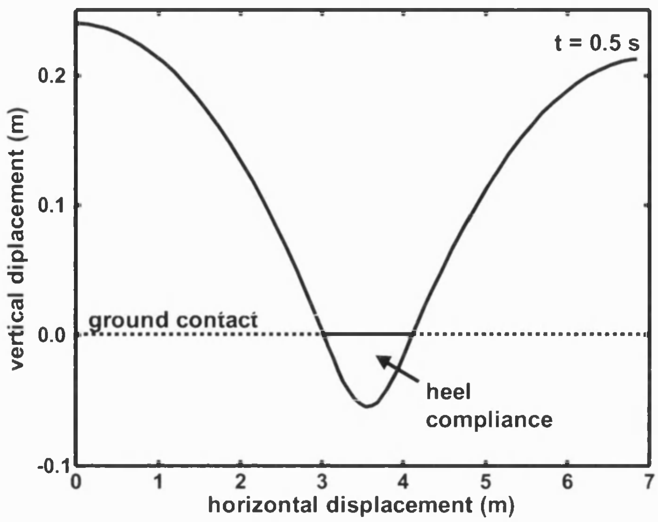


Figure 4.4 Horizontal versus vertical displacement of a contact step simulation with a spring-damper system in Simmechanics. Total duration of heel compliance is 0.14 s.

4.4.2. Conclusions

It was concluded that the spring-damper model created in Simmechanics was capable of creating a bouncing step movement. The size of the movement was however not realistic. To be able to use the spring-damper system in the trip recovery better estimation of spring and damper parameters are required and a combination of several spring-damper systems with non-linear instead of linear springs was expected to result in a more accurate simulation of foot-ground contact.

4.5. Alexander jumping model

None of the models described so far incorporated any force producing components representing the result of muscle action, such as joint moment generators. The trip recovery model will be joint moment (torque) driven. To gain insight and practice with a torque-driven model, a jumping model developed by Alexander (1990) was replicated in Simmechanics. The model consisted of a mass-less shank connected to the ground by a revolute joint, the ankle (Figure 4.5a). This shank was connected to a mass-less thigh with another revolute joint, the knee. This thigh was connected with a weld joint to the upper body, containing all of the body mass. The model was given an initial horizontal velocity of the upper body and motion was then driven by a torque generator at the knee (Figure 4.5b). However this initial velocity could not be created within Simmechanics. An initial horizontal velocity could only be specified with an initial position. Giving the hip an initial position however would over-determine the model. To get round this problem a force was applied to the CM of the upper body for a short time, creating a horizontal velocity. A stick figure of the model is shown in Figure 4.5g. Simulations ran until the knee torque reached a value of 0, and at that moment a stop subsystem (Figure 4.5e) stopped the simulation. At this instant jump height and jump distance were calculated in separate subsystems (Figure 4.5c and d) using projectile motion equations.

4.5.1. Simulations

Simulations were performed without an initial horizontal velocity. A horizontal force was applied to the upper body to create a delayed initial velocity. Because of the delayed initial velocity the torque dropped to zero too fast and take off occurred too early, so only a small jumping height and distance were reached. For this reason the moment of take off was not defined by when the torque dropped below zero, but by when the expected jump height was reached. A high and a long jump were simulated. Values were taken from Alexander (1990); maximum torque = 858.38 Nm, initial torque = 515.03 Nm, maximum knee angular velocity = 35.44°/s, initial knee angle = 170° and initial hip-ground angle = 45°. The horizontal force and the time it was applied for were varied to get results similar to Alexander (1990). Results are shown in Table 4.2. For the simulation of the high jump a horizontal force of 462 kN was applied for 1 ms and take-off occurred after 1.9 ms. For the long jump a force of 700 kN was applied for 1 ms and take-off occurred at 1.1 ms.

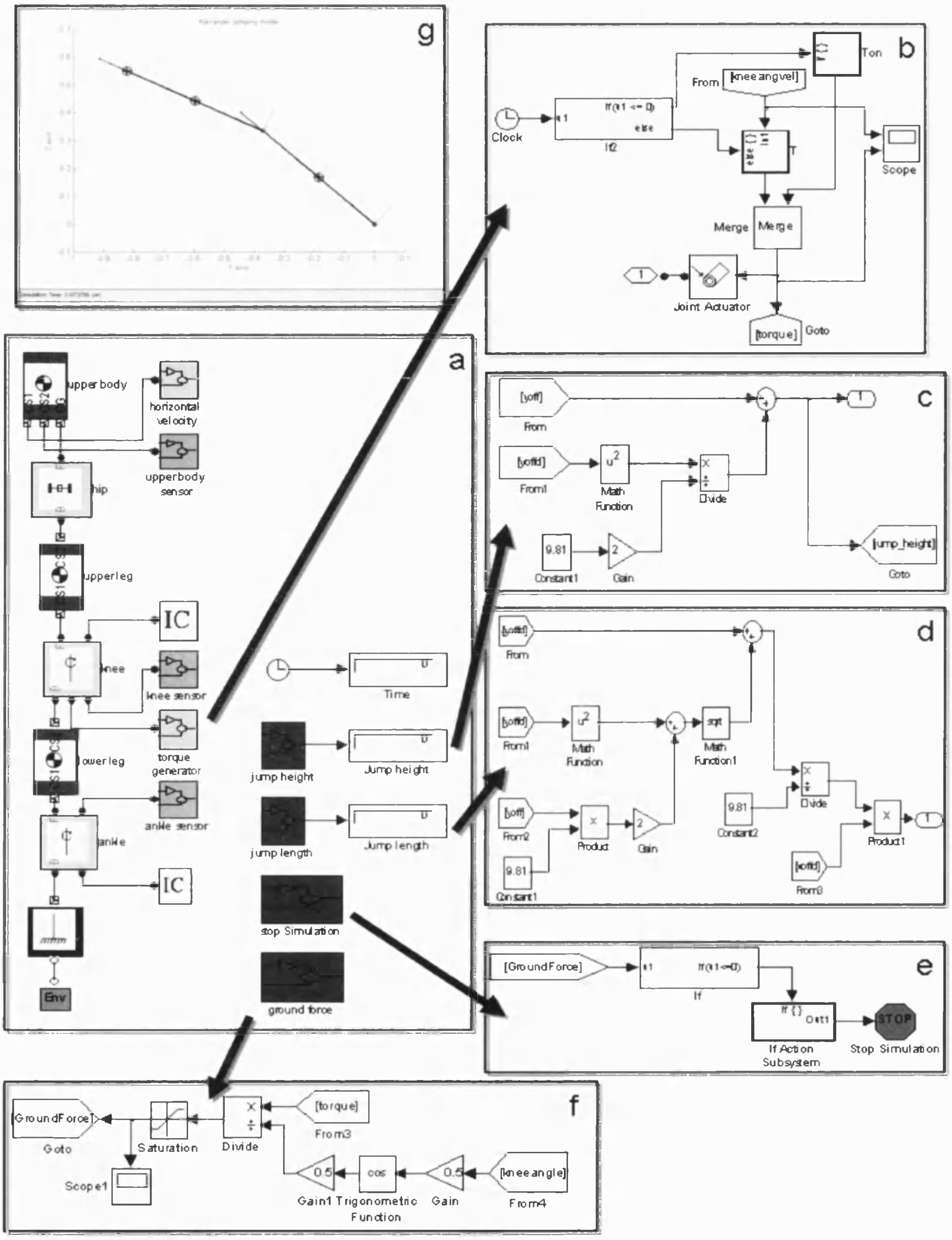


Figure 4.5 Architecture of the replicated jumping model, with (a) as the main model and (b) to (f) as subsystems, including torque generator (b), calculation of the jump distance (c), calculation of the jump height (d), stop criterion of the simulation (e) and the ground reaction force (f). A stick figure of the jumping model is shown in (g).

4.5.2. Conclusions

The simulations with Simmechanics obtained results close to Alexander's. The heigth of the high jump was estimated with an error of 4.5% and the length of the long jump with an error of 10.7%. Both values were overestimated. This was probably due to the application of the horizontal force instead of an initial horizontal velocity. The horizontal forces applied were high 462 kN and 700 kN. The time to take off was short in both simulations 1.9 ms and 1.1 ms, which might also be due by the high horizontal force. It is physically impossible for an athlete to create a ground reaction force large enough for take off in such a short time. It was therefore concluded that the Simmechanics model did not satisfactorily simulate the simple jumping model created by Alexander. The model did not have an initial horizontal velocity and did not remain within the physical boundaries of a human athlete. Because of the inability to provide any initial velocity to the body it was decided to produce the trip recovery model using Autolev instead of Simmechanics.

Table 4.2 Comparison of values from Alexander (1990) and simulated values.

	Alexander (1990)	Simulation
	High jump	
Jump height (m)	1.40	1.55
	Long jump	
Jump length (m)	8.00	8.36

4.6. Trip recovery model

A ten-segment torque driven trip recovery model was developed to aid the investigation of the contributions to successful trip recovery. The model had two feet each consisting of a rearfoot and a forefoot segment at a fixed angle to each other, two shank segments, two thigh segments and a pelvis segment. The upper body was represented by a single mass which could be moved relative to the pelvis segment. This allowed for the effect of arm movement without having to add upper body and arm segments to the model. The equations of motion were produced in Autolev (Appendix D) and the model was further written in C++. The model had three horizontal and three vertical spring-damper systems at each foot and six torque generators at the joints of the lower limbs. The torque profiles were modelled using a nine parameter function based on peak torque-angle-angular velocity relationships obtained from literature. Activation of the torque generators was controlled by a ramped activation function. The model simulated trip recovery from the swing phase preceding the trip, followed by perturbation of the swing leg, to the end of the initial recovery step.

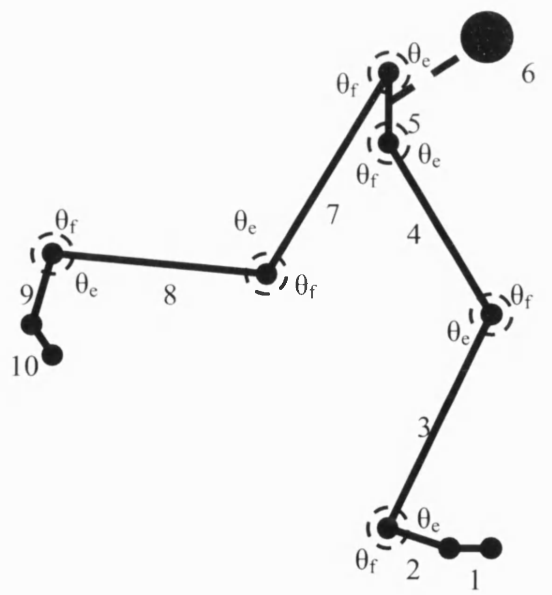


Figure 4.6 Representation of the ten-segment trip recovery model, including forefoot segments (1 and 10), rearfoot segments (2 and 9), shanks (3 and 8), thighs (4 and 7), pelvis (5), and upper body mass (6), θ_e: extensor torque generators at joints and θ_f: flexor torque generators at joints.

4.6.1. Ground contact

In reality foot-ground contact is complex with spring and damper properties in multiple directions spread over the foot and dependent on each other. In the simulation model this complex system was simplified. This required making several assumptions, which are described in this section.

Foot-ground contact was modelled with three horizontal and three vertical spring-damper systems at the ankle, metatarsal and toe of the foot segments, independent on each other. The foot segments were divided in two parts; rearfoot and forefoot, which remained at a fixed angle with each other (Figure 4.7). This allowed a certain amount of foot roll during ground contact.

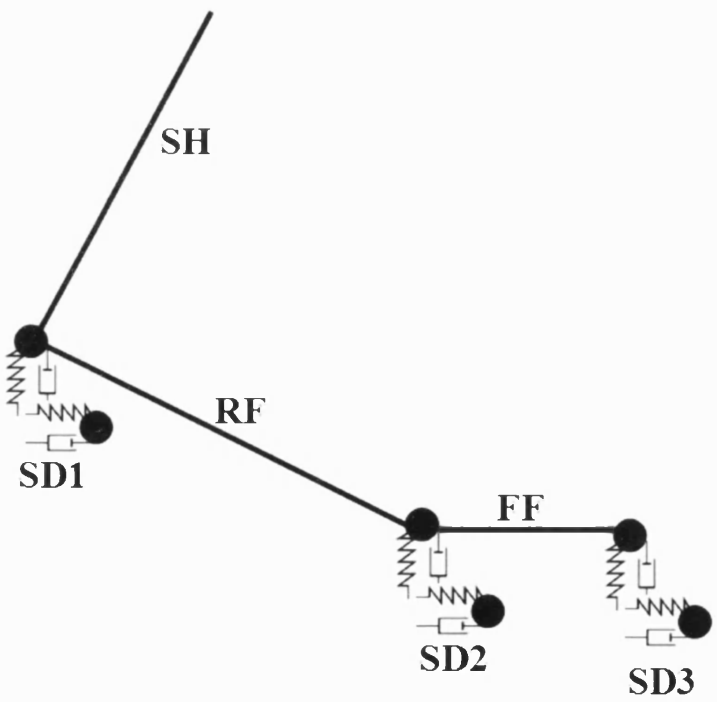


Figure 4.7 Foot segment with spring-damper segments at the ankle (SD1), metatarsal (SD2) and toe (SD3). SH is the shank segment, RF is the rearfoot segment and FF is the forefoot segment.

During ground contact spring-damper systems were initially set to be at their natural length, compress and return to their natural length at the end of ground contact. To account for continuous progression of the foot in the direction of the walk and to enable the springs-damper systems to return to their natural length some adjustments had to be made to the experimental displacements of the foot markers.

To model ground contact, estimates of the spring and damper constants were required that resulted in realistic simulations of the ground reaction forces. These were determined through matching experiments trip trials using the Downhill Simplex method. The Downhill Simplex method was chosen as it is robust and computationally efficient (Press et al., 2002). Based on optimisations by Gittoes (2004) and Wilson (2003) an RMS difference smaller than 25% of the maximum force was considered acceptable. The experimental data were collected with force plate (1000 Hz) and high speed video (1000 Hz) with the ankle, metatarsal and toe digitised in Peak Motus (as described in section 3.3). The simulated ground reaction force (F_{ysimi} and F_{zsimi}) was calculated with the measured displacements of relevant anatomical landmarks (y_{exp} and z_{exp}) and quadratic spring-damper equations (Equation 4.1 and Equation 4.2).

Equation 4.1

$$F_{ysim}(t) = (-k * y_{\exp}(t) * |y_{\exp}(t)| - r * y_{\exp}dt(t) * |y_{\exp}(t)| / (1 + c_y * |y_{\exp}(t)|)) * z_{\exp}(t)^2$$

Equation 4.2 $F_{zsim}(t) = (-k * z_{\exp}(t) - r * z_{\exp}dt(t)) * |z_{\exp}(t)|$

Where $y_{exp}(t)$ and $z_{exp}(t)$ are the horizontal and vertical displacements and $y_{exp}dt(t)$ and $z_{exp}dt(t)$ are the horizontal and vertical displacement velocities of the spring-damper system, k is the spring coefficient, r is the damper coefficient and c_y is a constant. To calculate the displacements of the spring-damper systems contact was considered as the natural length of the spring (zero displacement) and compression as the displacement from this point. For matching the experimental data to simulated data the high-speed video and force data were resampled from 1000 Hz to 200 Hz to reduce the influence of noise caused by digitising error.

Ground contact could be defined either by eye from the high speed video recordings or from the F_{zexpi} (measured vertical force) signal. The latter was done for the ankle and toe, metatarsal contact was based on observation of the high speed video data. Ground contact of the heel was assumed when F_{zexpi} was above 5.0 N and heel off was defined when the heel reached the same vertical position as on contact. Toe off was assumed when F_{zexpi}

dropped below 5.0 N, and toe contact was when the vertical toe position was equal to that at toe off.

During trip recovery the whole body and therefore the contact foot moved forward, this movement was relatively large compared to the spring-displacements that were required to calculate the ground reaction forces and dwarfed these spring-displacements. This made it impossible to measure the horizontal spring-damper displacements from raw motion data. To overcome this problem the horizontal displacements were calculated relative to the centre of mass (CM) of the foot. This also ensured the horizontal displacement at the ankle was in the opposite direction to that at the metatarsal and toe, which was necessary to simulate the inversion of F_{yexpi} when going from the touch down to the push off phase. At touch down only the ankle was in contact with the ground causing a negative F_{yexpi}. This negative F_{yexpi} reduced when the metatarsal and toe came in contact with the ground and finally became positive in the push off phase.

Because of the continuous forward movement of the foot the horizontal spring displacements did not return to its natural length at the end of ground contact. To overcome this problem the horizontal displacements (y_{exp}) were mirrored halfway through ground contact, forcing the displacements to return to their natural length and simulating the unloading of the springs (Figure 4.8).

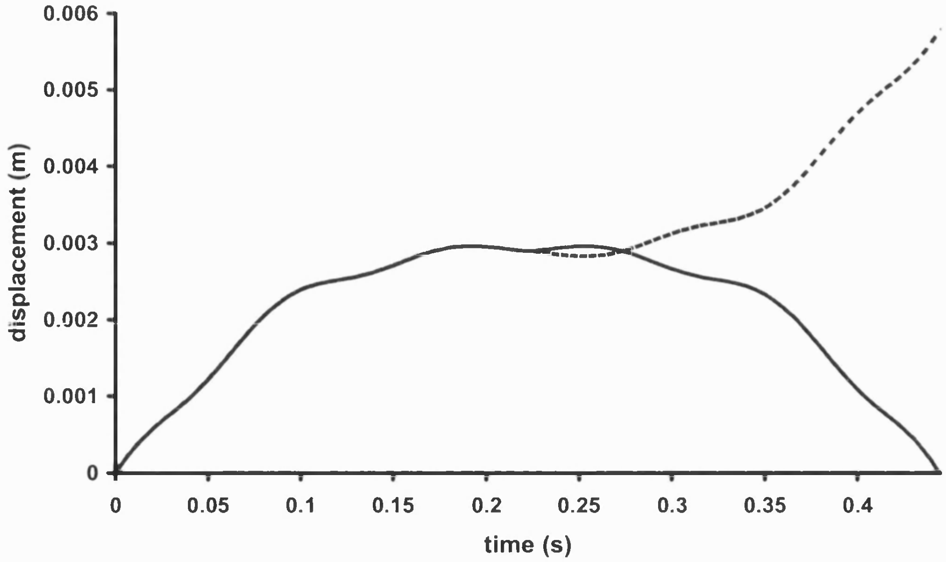


Figure 4.8 Horizontal displacement of the ankle marker during an elevating strategy recovery. The marker displacement is inverted halfway during contact to represent spring loading and unloading. The dotted line is the marker displacement and the solid line the spring displacement.

The RMS differences between the simulated forces and the experimental forces ($Score_y$ and $Score_z$) were calculated, as a percentage of the maximum force ($F_{ymaxexp}$ and $F_{zmaxexp}$). $Score_y$ and $Score_z$ were used as objective functions in the Downhill Simplex routine to optimise the spring and damper parameters (k and r) for the horizontal and vertical springs at the ankle, metatarsal and toe.

Spring and damper parameters were optimised for seven trials of an older adult and three trials of a younger adult (Table 4.3, Table 4.4 and Table 4.5). As there was a large variation in the spring and damper parameters between the trials, it was chosen to use the parameters from a single trial in the trip recovery simulation model. It was chosen not to use mean values as it was assumed that the ratio of the variables to each other were important in simulating the ground reaction forces. The trials with the lowest $Score_y$ and $Score_z$ were used; this was trial 1 for the older subject and trial 2 for the younger subject.

This resulted in acceptable results for the simulated F_y with an RMS difference between simulated and measured force of 11.0% for a lowering strategy recovery trial of an older adult (Figure 4.9). F_z was simulated with an acceptable RMS difference between simulated and measured results below 5.8% for an older adult (Figure 4.10).

Table 4.3 Optimised spring and damper parameters for the ankle with $Score_y$ and $Score_z$ values for seven trials of an older adult and three trials of a younger adult. And with k_y the horizontal spring parameter, k_z the vertical spring parameter, r_y the horizontal damper parameter and r_z the vertical damper parameter.

	#	k_y (Nm^{-1})	k_z (Nm^{-1})	r_y (Nsm^{-1})	r_z (Nsm^{-1})	$Score_y$ (%)	$Score_z$ (%)
Older	1	$317*10^9$	$533*10^3$	$138*10^6$	$98*10^3$	11.0	5.8
	2	$3299*10^6$	$1218*10^3$	$7469*10^3$	5772	33.3	13.0
	3	$11*10^9$	$3270*10^3$	$4631*10^3$	$98*10^3$	12.7	7.7
	4	$21*10^9$	$2202*10^3$	$4095*10^3$	$37*10^3$	5.9	12.4
	5	$40*10^9$	$425*10^3$	$18*10^6$	$303*10^3$	60.8	18.6
	6	$58*10^9$	$1471*10^3$	$132*10^6$	2871	17.6	7.3
	7	$59*10^6$	$756*10^3$	$332*10^6$	$13*10^3$	19.6	16.8
Younger	1	$925*10^9$	$684*10^3$	$40*10^6$	$126*10^3$	49.6	32.4
	2	$6333*10^6$	$525*10^3$	$14*10^6$	$1828*10^3$	18.7	7.0
	3	$3258*10^6$	$855*10^3$	$58*10^6$	$626*10^3$	41.5	12.1

Table 4.4 Optimised spring and damper parameters for the metatarsal with $Score_y$ *and* $Score_z$ *values for seven trials of an older adult and three trials of a younger adult. And with* k_y *the horizontal spring parameter,* k_z *the vertical spring parameter,* r_y *the horizontal damper parameter and* r_z *the vertical damper parameter.*

	#	k_y (Nm^{-1})	k_z (Nm^{-1})	r_y (Nsm^{-1})	r_z (Nsm^{-1})	$Score_y$ (%)	$Score_z$ (%)
Older	1	$1044*10^6$	$6963*10^3$	$177*10^3$	$121*10^3$	11.0	5.8
	2	$418*10^9$	$1041*10^3$	$55*10^6$	1959	33.3	13.0
	3	$35*10^6$	$3280*10^3$	5962	$19*10^3$	12.7	7.7
	4	$670*10^6$	$502*10^3$	$1145*10^3$	0	5.9	12.4
	5	$132*10^6$	$776*10^3$	$23*10^3$	$627*10^3$	60.8	18.6
	6	$739*10^6$	$2728*10^3$	$96*10^3$	1941	17.6	7.3
	7	$43*10^3$	$172*10^3$	$332*10^6$	0	19.6	16.8
Younger	1	$3035*10^6$	$8935*10^3$	$518*10^3$	$156*10^3$	49.6	32.4
	2	$80*10^6$	$1828*10^3$	$10*10^3$	$27*10^3$	18.7	7.0
	3	$33*10^6$	$626*10^3$	$581*10^3$	596	41.5	12.1

Table 4.5 Optimised spring and damper parameters for the toe with $Score_y$ and $Score_z$ values for seven trials of an older adult and three trials of a younger adult. And with k_y the horizontal spring parameter, k_z the vertical spring parameter, r_y the horizontal damper parameter and r_z the vertical damper parameter.

	#	k_y (Nm^{-1})	k_z (Nm^{-1})	r_y (Nsm^{-1})	r_z (Nsm^{-1})	$Score_y$ (%)	$Score_z$ (%)
Older	1	$1044*10^6$	$678*10^3$	$177*10^3$	$12*10^3$	11.0	5.8
	2	$418*10^9$	$13*10^6$	$54*10^6$	$20*10^3$	33.3	13.0
	3	$35*10^6$	$660*10^3$	5962	1725	12.7	7.7
	4	$670*10^6$	$48*10^3$	$1145*10^3$	207	5.9	12.4
	5	$13*10^9$	$1192*10^3$	$23*10^3$	$18*10^3$	60.8	18.6
	6	$739*10^6$	$878*10^3$	$96*10^3$	6850	17.6	7.3
	7	$43*10^3$	$17*10^3$	$43*10^3$	71	19.6	16.8
Younger	1	$3035*10^6$	$870*10^3$	$518*10^3$	$16*10^3$	49.6	32.4
	2	$80*10^6$	$790*10^3$	$10*10^3$	$64*10^3$	18.7	7.0
	3	$33*10^6$	$1372*10^3$	$581*10^3$	915	41.5	12.1

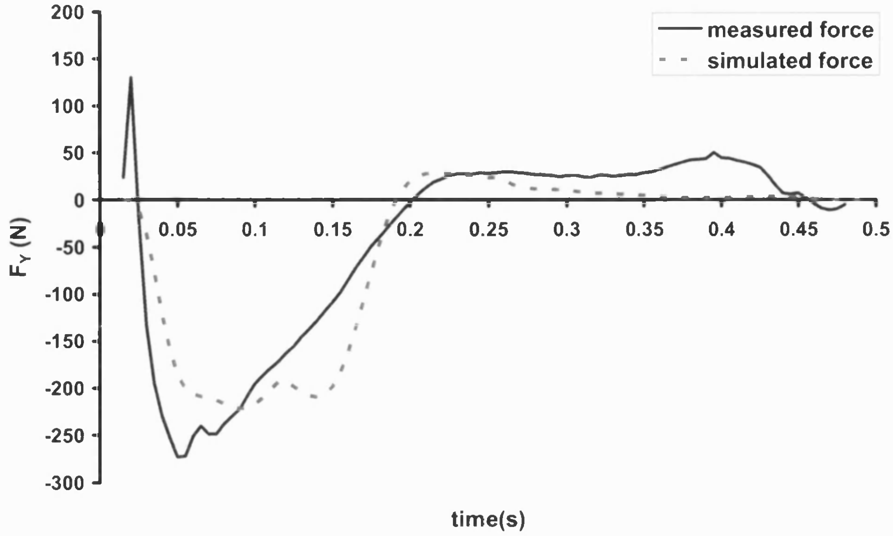


Figure 4.9 F_y calculated with horizontal displacement relative to the foot CM and optimised spring and damper constants (dotted line) and measured F_y (solid line).

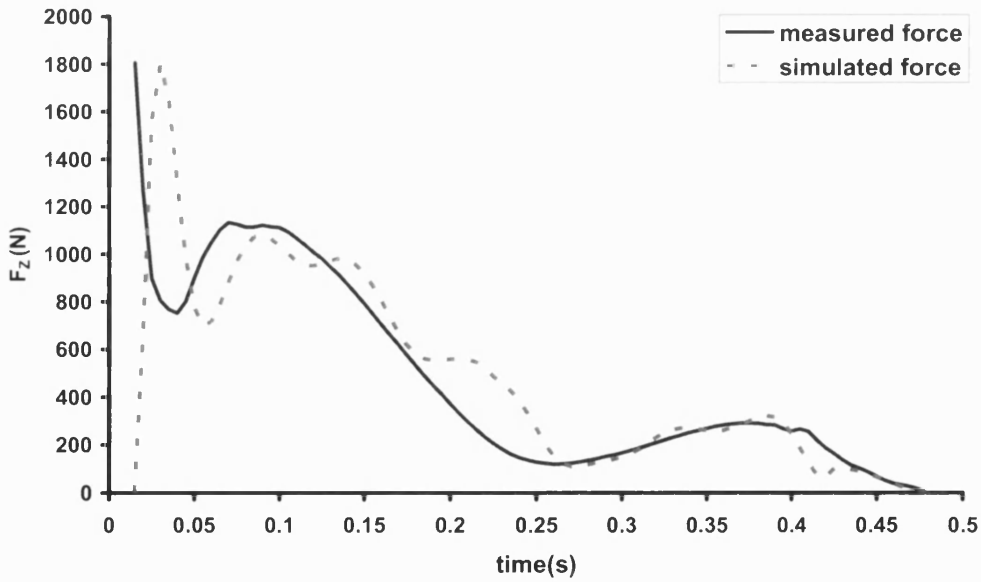


Figure 4.10 F_z simulated (dotted line) and measured F_z (solid line).

The simulation started during a walking stride and therefore started with one foot in contact with the ground. As a result of this the spring displacement at the start of the stride was unknown making it impossible to invert the horizontal displacement, which was needed to calculate F_y. For this reason a cosine function was developed that prescribed the vertical spring displacement (Equation 4.3). This function needed the lengths of the horizontal and vertical springs (ly_{spring} and lz_{spring}) and the initial position of the vertical spring (pz_{spring}). It was chosen to use a cosine function to initially calculate vertical spring displacement as this was the mathematically simplest function that provided similar displacements to the experimental ones, and which had a derivative that was similar to that of the displacement velocity data. As the displacement decreases following Equation 4.3 the simulation had to start at the unloading phase of the spring which was at mid-stance or later.

Equation 4.3 $$y(t) = 0.5 * y_{ini} * \cos\left(t * \frac{\pi}{t_{contactestimated}}\right) + 0.5 * y_{ini}$$

where $y_{ini} = \dfrac{ly_{spring} * pz_{spring}}{lz_{spring}}$

Where $y(t)$ is the horizontal spring displacement at time t, y_{ini} is the estimated horizontal displacement at the start of the simulation and $t_{contactestimated}$ is the estimated duration of ground contact of the spring-damper system. The ground contact time was unknown prior to the simulation and was estimated by repeating simulation of the ground contact phase until $t_{contactestimated}$ was within a certain limit of the actual contact duration.

During foot contact the horizontal displacements were inverted by storing displacements during the first half of contact and inverting these stored values during the second half. The duration of ground contact was again estimated by repeated loops, starting with the experimental duration of ground contact of the recovery step.

4.6.2. Joint torques

The flexor and extensor torque generators each existed of a contractile and an elastic element in series as described by Wilson (2003) to conform to the angular equivalent of the Hill muscle model (Hill, 1938).

The contractile component angle (θ_{cc}) and the elastic component angle (θ_{ee}) were calculated in such a way that for extensors the internal angle of the joint (θ) was equal to 2π minus the sum of θ_{cc} and θ_{ee} and for flexors the internal angle of the joint was equal to the sum of θ_{cc} and θ_{ee} (Figure 4.11).

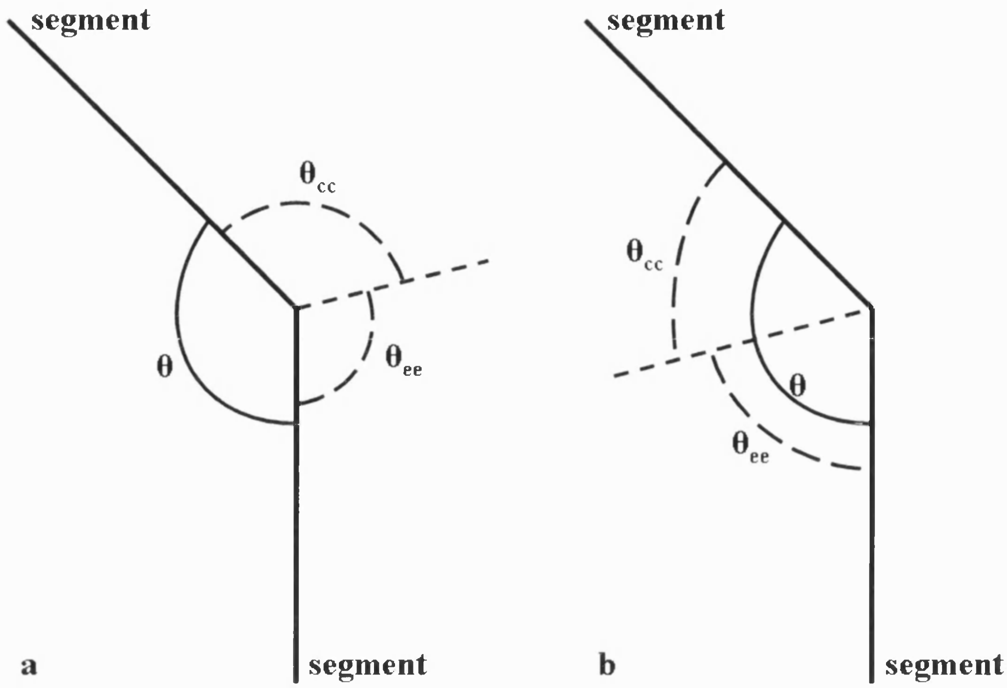


Figure 4.11 (a) Contractile (θ_{cc}) and elastic (θ_{ee}) component angle of an extensor torque generator and (b) contractile and elastic component angle of a flexor torque generator.

The torque at the elastic elements were calculated as a rotational spring with θ_{ee} the elastic component angle and K_e the series elastic stiffness parameter (Wilson, 2003) (Equation 4.4).

Equation 4.4 $T_{ee} = K_e * \theta_{ee}$

The contractile element torques were calculated with a nine parameter function. This nine parameter function consisted of a four parameter function (Equation 4.5 and Equation 4.6) which expressed the maximum torque at full activation as a function of joint angular velocity (Yeadon et al., 2006). This four parameter function consisted of two hyperbolic functions representing the concentric and the eccentric phase. The first is equivalent to the Hill hyperbola (Hill, 1938). The four parameters defining the four parameter function were: T_{max}, the maximum torque in the eccentric phase, T_0, the isometric torque, ω_{max} the maximum angular velocity, and ω_c, the vertical asymptote of the Hill hyperbola (Figure 4.12). T_{max} is assumed to be 1.5 times T_0.

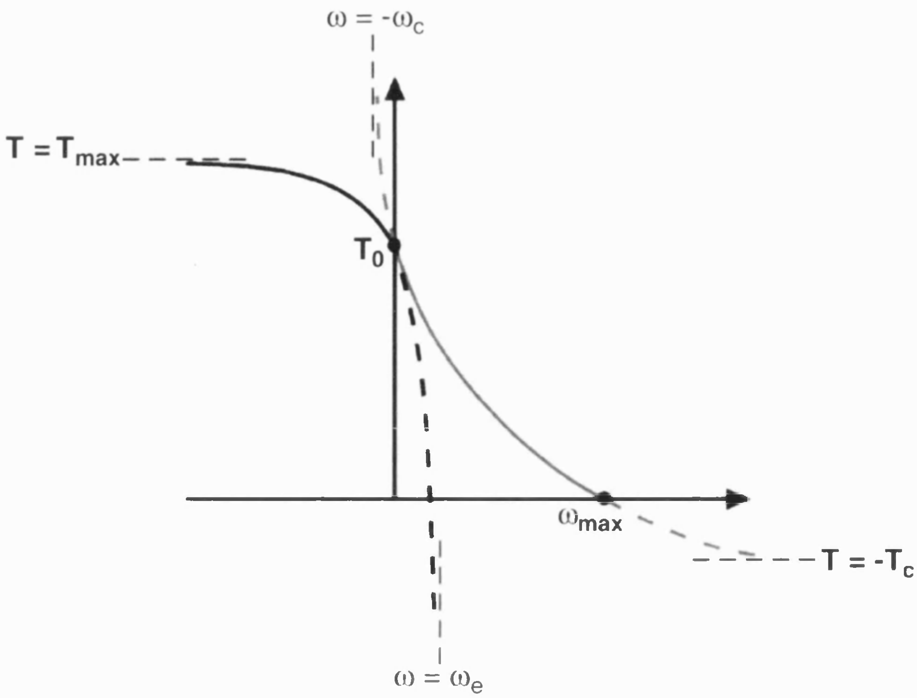


Figure 4.12 The four parameter torque function with T_{max}, the maximum torque in the eccentric phase, T_0, the isometric torque, ω_{max} the maximum angular velocity, and ω_c, the vertical asymptote of the Hill hyperbola (Yeadon et al., 2006).

Equation 4.5 $T = \dfrac{C}{(\omega_c + \omega)} - T_c \qquad (if\, \omega \geq 0),$

where $T_c = \dfrac{T_0 * \omega_c}{\omega_{\max}}$, $C = T_c * (\omega_{\max} + \omega_c)$

Equation 4.6 $T = \dfrac{E}{(\omega_e - \omega)} + T_{\max} \qquad (if\, \omega \leq 0)$

where $\omega_e = \dfrac{(T_{\max} - T_0)}{kT_0} - \dfrac{\omega_{\max} * \omega_c}{(\omega_{\max} + \omega_c)}$, $E = -(T_{\max} - T_0) * \omega_e$

The value of *k*, which represents the slopes of the eccentric and concentric functions at $\omega = 0$, was set to 4.3, which is the theoretical value predicted by Huxley (1957).

This four parameter function was multiplied by a differential activation, which was calculated with a three parameter function (Equation 4.7 and Figure 4.13). This function corrected for the fact that full activation is not achieved in eccentric contractions

(Wresling, 1990). The three parameters in the function were a_{min}, minimal activation, ω_1, the angular velocity at the point of inflection of the function and m, which governs the rate at which the activation increases with angular velocity. The maximum activation (a_{max}) was assumed to be 1.0.

Equation 4.7 $$\omega - \omega_1 = \frac{+m * \left(a - 0.5 * \left(a_{min} + a_{max}\right)\right)}{\left(a_{max} - a\right)\left(a - a_{min}\right)}$$

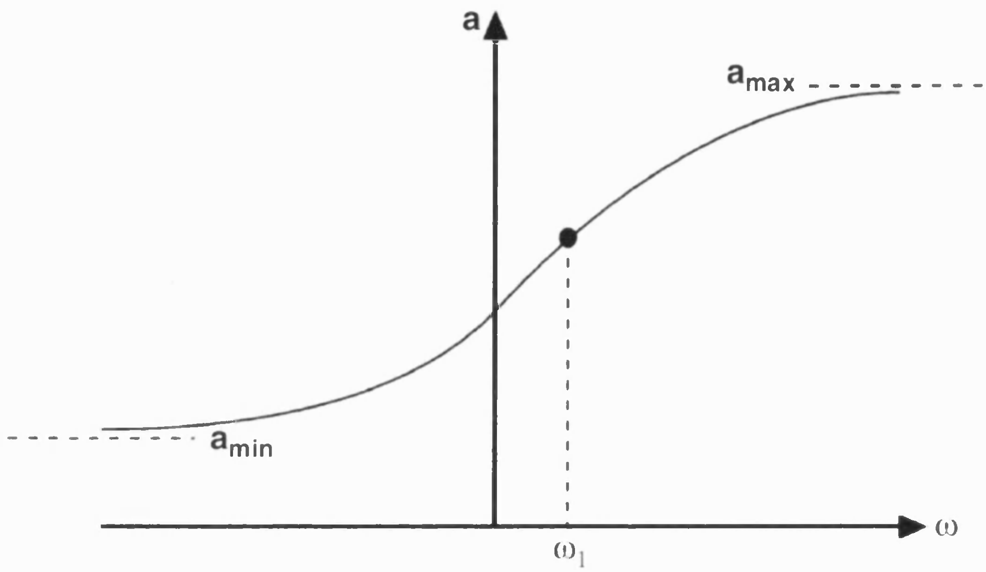


Figure 4.13 The differential activation function with a_{min}, the minimum activation level, a_{max}, the maximum activation level and ω_1 the inflection point of the equation (Yeadon et al., 2006).

The seven parameter function was finally multiplied with a two parameter function (Equation 4.8) to make the torque angle-dependent.

Equation 4.8 $$T_{(\theta,\omega)} = T_{\omega} * \left(1 - k_2 * \left(\theta - \theta_{opt}\right)^2\right)$$

with T_{ω}: the seven parameter function, k_2: the rate at which the torque drops off from the optimum angle and θ_{opt}: the optimum angle at which maximum torque occurs.

Contractile component parameters

With this nine parameter function, six torque-angle-angular velocity plots were produced for both younger and older adults; for ankle dorsi flexor and plantar flexor and for knee and hip extensor and flexor torques. The values for the nine parameters were all group-specific and would ideally be obtained by fitting the nine parameter function to experimental isokinetic dynamometer data. It was however decided that dynamometer measurements would be too strenuous for the older subject group in addition to the trip recovery experiment. Therefore the nine parameters for the torque function were approximated using values from literature for female subjects of similar age. While acknowledging the limitations of this approach, estimating the parameters from literature was considered permissible as the model is a simplification of reality in any case and cannot exactly replicate it. At first the seven parameters were calculated by optimising the seven parameter function to the data from literature. For this optimisation the RMS difference (*Score*) was calculated between the torque values from literature (lit_i) and the calculated torque values (sim_i). The value of *Score* was the objective function in the optimisation of the seven parameters for the torque function with the Downhill Simplex method. No constraints were used in this optimisation.

The optimised activation function, four parameter function and seven parameter function for ankle plantar flexion of older women are given in Figure 4.14, Figure 4.15 and Figure 4.16 respectively.

The final two parameters of the nine parameter function were obtained by optimising the nine parameter function with Downhill Simplex in a similar way to optimisation of the seven parameters. The previously optimised seven parameters and torque-angle profiles at different angular velocities from literature were used in the optimisation of these final two parameters. The estimated torque-angle-angular velocity profiles for extensors and flexors of the ankle, knee and hip of both older and younger women are shown in Figure 4.17 to Figure 4.22 and the agreeing nine parameters in Table 4.6.

There was limited literature specifying complete torque-angle-angular velocity relationship so some torque profiles were estimated from data on different torque generators or age groups. No values could be found in literature to create ankle dorsi flexor torque-angle-angular velocity profiles. These profiles were therefore calculated relative to the plantar flexor torque based on values described by Wilson (2003). Wilson (2003) measured torque on a single male athlete. Values for ω_{max}, ω_c, a_{min}, m and ω_1 are kept the same as for ankle plantar flexion. Plantar flexor values for T_{max} were multiplied by 0.226, values for T_0 by 0.231, values for θ_{opt} by 1.267 and values for k_2 by 1.119 to get the dorsi flexor torque profile. For the knee extensors and flexors no data were available in literature to describe the torque-angle relationship, this relationship was therefore based on k_2 and θ_{opt} by Wilson (2003). For knee flexor torques of younger women only limited torque-angular velocity relationship data could be found. Therefore a profile was created relative to the knee extensor torque profile based on the knee flexor-extensor ratio in Wilson (2003) averaged with a profile fitted to data from Arnold et al. (1997). A nine parameter function was fitted to this average profile. The ratios of the parameters for the profile relative to the extensor profile were 0.656 for T_0, 0.778 for ω_c, 1.119 for ω_{max}, 1.036 for a_{min}, 0.878 for m and 0.482 for ω_1. As no torque-angle or torque-angular velocity relationships were found in literature for the hip, the torque profiles for the hip were calculated by averaging seven parameter function from the ankle and knee and shifting this to T_0 values by Dean et al. (2004). T_0 for hip flexion was 88.8 Nm for older and 106.7 Nm for younger women, for hip extension T_0 was 88.2 Nm for older and 108.1 Nm for younger women. Values for k_2 and θ_{opt} were again taken from Wilson (2003).

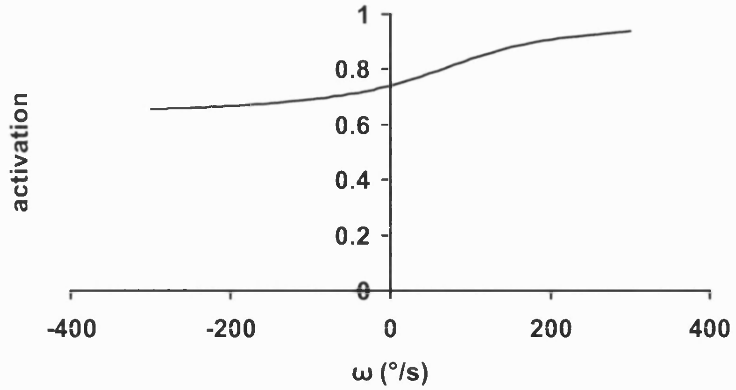


Figure 4.14 Differential activation for ankle plantar flexor.

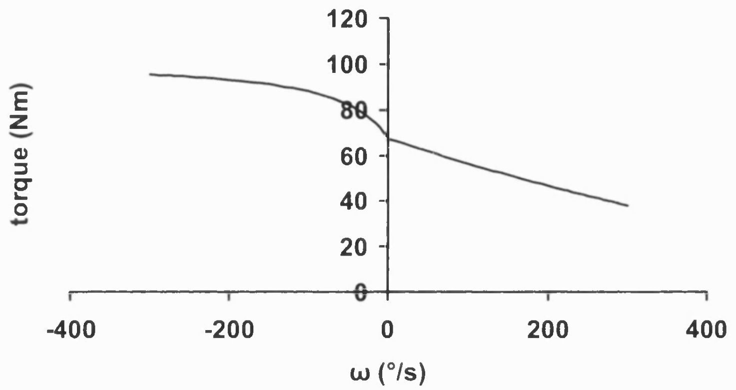


Figure 4.15 Four parameter fit for ankle plantar flexor torque-angular velocity relationship.

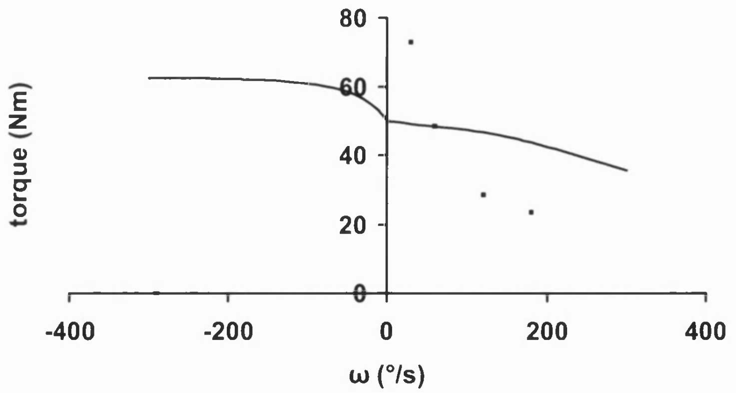


Figure 4.16 Seven parameter fit of ankle plantar flexor torque-angular velocity relationship (solid line) with values from Gajdosik et al. (1999).

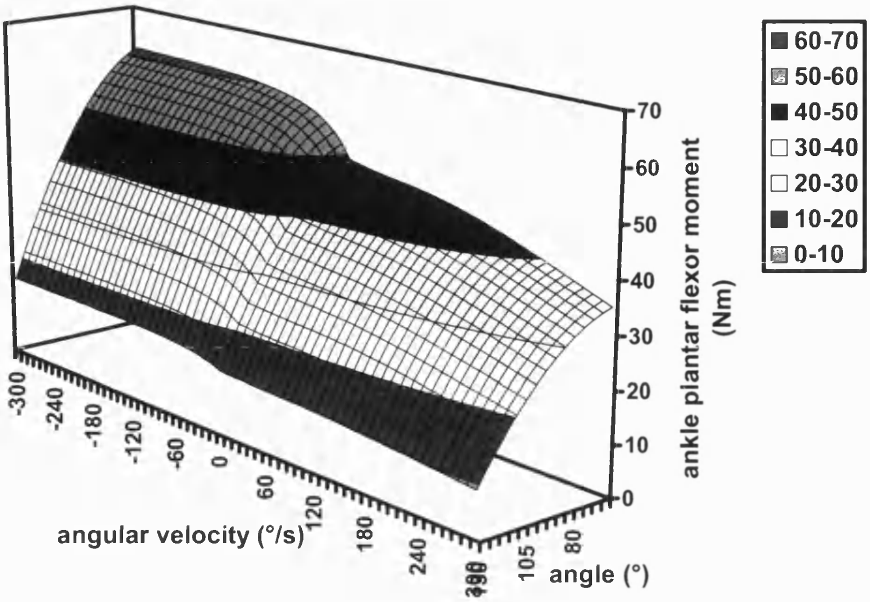


Figure 4.17 Surface plot of the torque-angle-angular velocity relationship for ankle plantar flexors of older women, based on values described by Gajdosik et al. (1999).

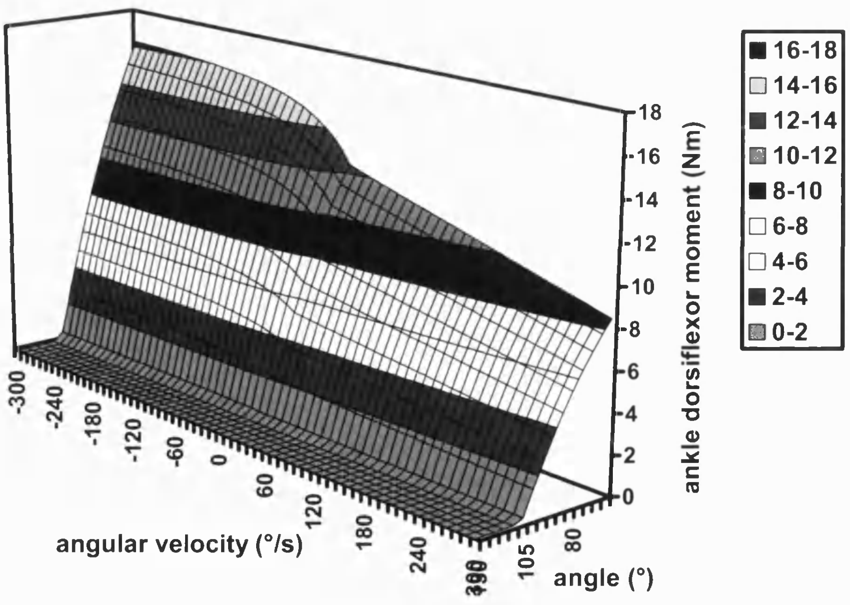


Figure 4.18 Surface plot of the torque-angle-angular velocity relationship for ankle dorsi flexors of older women.

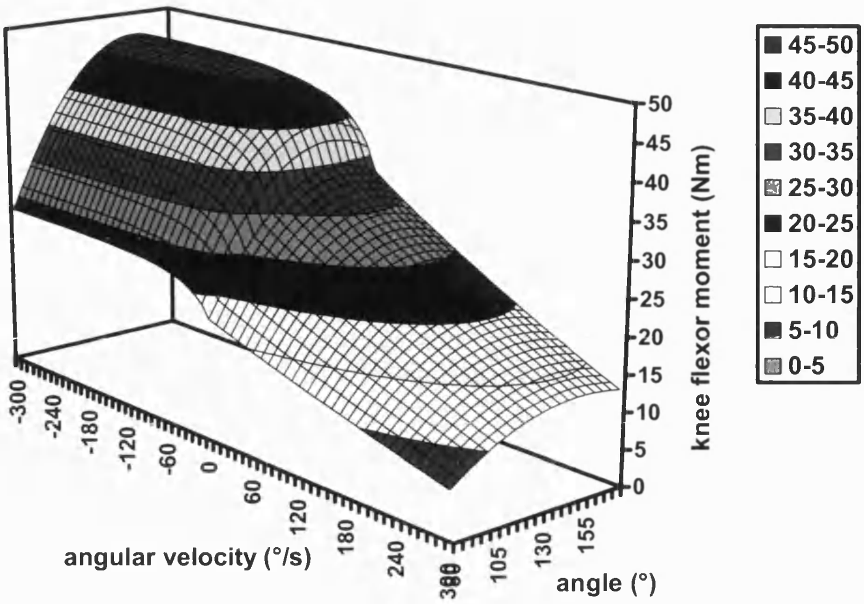


Figure 4.19 Surface plot of the torque-angle-angular velocity relationship for knee flexors of older women, based on data by Capranica et al. (1998) and Wilson (2003).

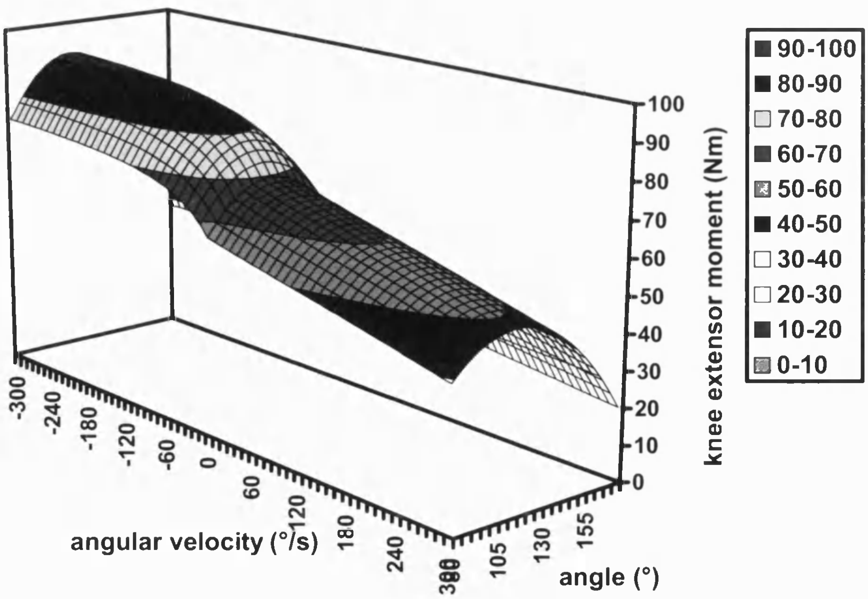


Figure 4.20 Surface plot of the torque-angle-angular velocity relationship for knee extensors of older women, based on values by Capranica et al. (1998), Bellew and Yates (2000) and Wilson (2003).

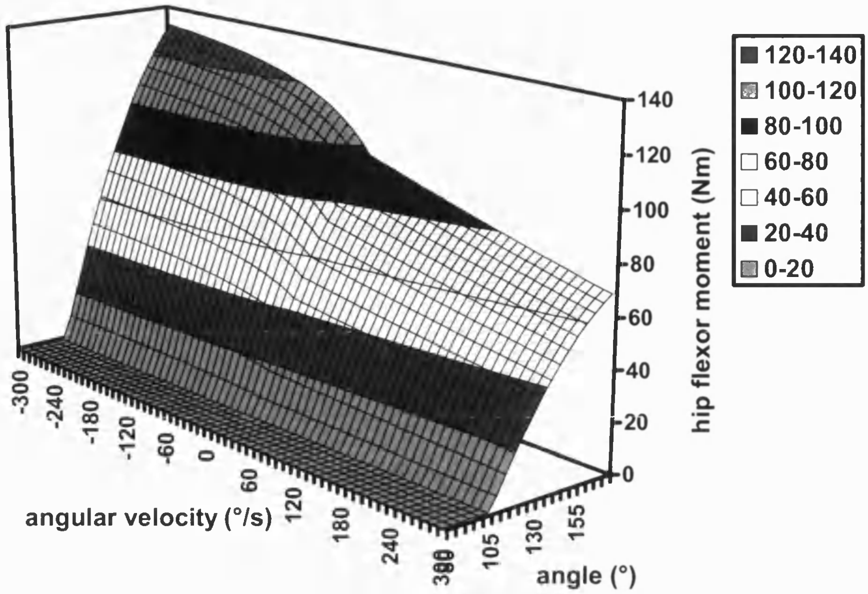


Figure 4.21 Surface plot of the torque-angle-angular velocity relationship for hip flexors of older women, based on values by Dean et al. (2004).

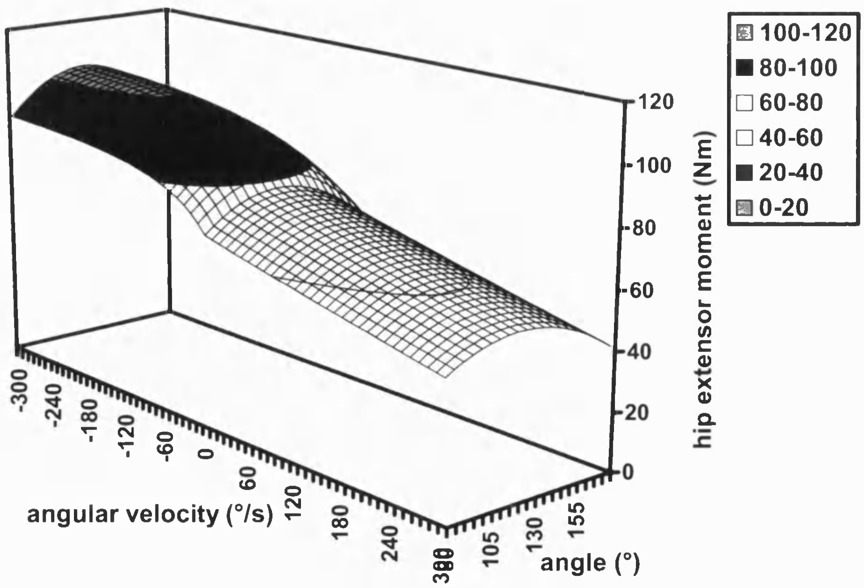


Figure 4.22 Surface plot of the torque-angle-angular velocity relationship for hip extensors of older women, based on values by Dean et al. (2004).

Table 4.6 Optimised parameters for nine parameter function, for older and younger women. With DF as dorsi flexion, PF as plantar flexion, Ex as extension and Fl as flexion.

		Ankle DF	Ankle PF	Knee Ex	Knee Fl	Hip Ex	Hip Fl
Younger women	T_0 (Nm)	18.7	81.1	117.1	85.8	108.1	103.7
	T_{max} (Nm)	27.5	121.7	175.7	128.7	162.2	155.6
	ω_c (rad/s)	29.5	29.5	42.6	31.2	39.3	37.7
	ω_{max} (rad/s)	19.8	19.8	28.5	20.9	26.3	25.3
	a_{min}	0.738	0.738	1.065	0.780	0.983	0.943
	m	0.72	0.72	1.04	0.76	0.96	0.92
	ω_1 (rad/s)	1.48	1.48	2.13	1.56	1.97	1.89
	θ_{opt} (rad)	2.95	2.33	2.00	2.13	1.6	3.6
	k_2	0.42	0.35	0.53	0.32	0.27	0.33
Older women	T_0 (Nm)	15.6	67.6	82.5	41.1	88.2	88.9
	T_{max} (Nm)	22.9	101.4	123.7	61.6	132.3	133.4
	ω_c (d/s)	24.6	24.6	30.0	14.9	32.1	32.3
	ω_{max} (rad/s)	16.5	16.5	20.1	10.0	21.5	21.7
	a_{min}	0.615	0.615	0.750	0.750	0.802	0.809
	m	0.60	0.60	0.73	0.73	0.79	0.79
	ω_1 (rad/s)	1.23	1.23	1.50	1.50	1.60	1.62
	θ_{opt} (rad)	2.95	2.33	2.00	2.13	1.6	3.6
	k_2	0.42	0.35	0.53	0.32	0.27	0.33

Elastic component parameters

The torque produced by the series elastic components was dependent on the angle of the series elastic component and the series elastic component stiffness of each joint. This series elastic component stiffness needed to be estimated for flexion and extension of each joint. The series elastic component was represented by the tendon and parts of the muscle (Wilson, 2003). Properties of the muscles and tendons were estimated from literature (Table 4.7) (Pierrynowski, 1995; Jacobs et al., 1996; Maganaris et al., 1998, 2000; Narici et al., 2003; Wilson, 2003; Maganaris, 2004; Morse et al., 2004; Mian et al., 2007); where possible data on older adults were used. The length of the series elastic component, which was required to calculate the series elastic component stiffness, was calculated using data from literature on muscle belly length, individual fibre lengths and pennation angles of older adults (Narici et al., 2003; Morse et al., 2004; Karamanidis & Arampatzis, 2006). The length over which the aponeurosis extends along the muscle belly was calculated by multiplying individual fibre length by the cosine of the pennation angle (Wilson, 2003). The maximum isometric torque, as estimated in the previous section, was distributed over the muscle groups acting at that joint according to the ratio of cross-sectional area times moment arm.

Tendon stiffness was determined for each muscle group. The change of length of the series elastic component was calculated using a percentage stretch of 5% based on literature values for isometric contractions (Wilson, 2003). This change in length was converted to a change in angle of the series elastic component by dividing it by the moment arm. The series elastic component stiffness was calculated by dividing the maximum isometric torque by the change in angle of the series elastic component during isometric contractions and then summing the stiffness values of the muscle groups at each joint (Table 4.8).

Table 4.7 Muscle and tendon properties scaled to body mass and height of an older participant.

	Series elastic component length (mm)	Muscle cross-sectional area (mm^2)	Moment arm (mm)	Maximum isometric joint torque (Nm)
Soleus	255	13939	48	54.7
Gastrocnemius (ankle)	316	3377	51	13.9
Tibialis anterior	315	2396	38	15.6
Rectus femoris (knee)	324	3943	41	28.1
Vastus lateralis	205	8081	39	54.4
Hamstrings (knee)	289	4684	24	28.9
Gluteus maximus	316	3377	14	12.1
Hamstrings (hip)	183	2333	58	25.4
Rectus femoris (hip)	289	4684	72	62.9

Table 4.8 Estimated series elastic component stiffness for each joint for an older female.

Torque generator	Series elastic component stiffness (Nm/rad)
Ankle plantar flexor	246.1
Ankle dorsi flexor	37.7
Knee flexor	59.7
Knee extensor	279.7
Hip flexor	178.9
Hip extensor	472.0

Muscle-tendon complex

Each torque generator consisted of a contractile and an elastic component in series. The definition of the contractile component, elastic component and joint angles for flexion and extension are shown in Figure 4.11. The nine parameter function needed to be converted to be in terms of contractile component joint angles and angular velocities rather than joint angles and angular velocities. The joint angles related to the elastic and contractile component angles as follows:

Equation 4.9 $2\pi = \theta + \theta_{cc} + \theta_{ee}$ (Extension)

Equation 4.10 $\theta = \theta_{cc} + \theta_{ee}$ (Flexion)

As the k_2 and θ_{opt} defined the torque-angle relationship in the nine parameter function these needed to be recalculated to describe the contractile component angle-torque relationship. This was done by setting series elastic component torque equal to contractile component torque calculated with the nine parameter function and calculating the corresponding θ_{ee} for each torque-angle-angular velocity relationship (Wilson, 2003). θ_{cc} could be calculated from θ_{ee} and was used to recalculate k_2 and θ_{opt} in the same manner they had been calculated initially. The recalculated values are shown in Table 4.9.

Table 4.9 Recalculated k_2 and θ_{opt} values.

Torque generator	k_2	θ_{opt}
Ankle plantar flexor	0.38	4.5
Ankle dorsi flexor	0.08	2.0
Knee flexor	0.26	4.0
Knee extensor	0.32	4.0
Hip flexor	0.16	1.0
Hip extensor	0.28	2.5

During model simulations the contractile and elastic element angles needed to be determined. This was done by assuming the torque at the elastic element was equal to that at the contractile component. Initially it was assumed that the velocity of the elastic component was zero and the angular velocity of the contractile component was therefore equal to the joint angular velocity (Wilson, 2003). The angles of the contractile and elastic components were calculated by assuming their torques were equal. The contractile component torque at the end of this initial iteration was calculated with the following equation:

Equation 4.11 $\theta_{cc} = \theta_{cci} + 0.5 * (\theta dt_{cc} + \theta dt_{cci})dt$

with θ_{cci} and θdt_{cci} the estimates of the contractile component angle and angular velocity at the beginning of the iteration. For each iteration after this initial iteration the contractile component angle was calculated with Equation 4.11 and the angular velocity of the contractile element needed to be calculated (Wilson, 2003). This velocity was calculated by assuming the torques of the elastic and contractile element were equal to each other. An iterative method was used to find the velocity for which the contractile and elastic elements were within 0.01 Nm of each other.

Passive torques

During simulations of trip recovery the movement of the ankle tended to move close to and outside its working range of motion. As the stiffness of a joint increases when it reaches the limits of its working range of motion it was decided to apply a passive torque at the ankle plantar flexors. This passive torque ensured the joint angle stayed within its physical possible range of motion. This passive torque linearly increased from 0 Nm at 131.2° to 67.6 Nm at 177.9°. This was based on literature on passive stretching and passive torque of the ankle joint in older adults (Gajdosik et al., 1996; Gajdosik, 1997; Mecagni et al., 2000; Gajdosik, 2002; Gajdosik et al., 2005). The maximum passive torque was set equal to the maximum isometric torque as passive torque at the dorsi flexors in elderly women was found to be close to the isometric torque used in the 9 parameter function.

4.6.3. Validation of torque profiles

The torque profiles described in the previous section were derived from values in literature. There was however a high variability in the data sourced. To get an indication of how well the joint torques of the participants in this study agreed with the calculated torque profiles maximum torque values were measured for one of the younger participants. This was the younger participant who was used to obtain input data for the trip recovery model evaluation.

Maximum joint torques of the ankles, knees and hips were measured on a Biodex (Multi Joint System 3 Pro, Nr. 830-200) dynamometer, at the full range of motion at three angular velocities (60, 120 and 180 °/s). Concentric torques were measured with an isovelocity concentric-concentric flexion-extension profile, and eccentric torques with an isovelocity eccentric-eccentric extension-flexion profile. Eccentric torques were not measured to the full limits of the joint range of motion, as at these limits it was impossible to produce enough force to trigger the dynamometer. The initial joint angle was measured with a goniometer and this information was used to calculate the joint angle from the crank angle. The mass of the limb and the crank arm of the dynamometer were weighed by the dynamometer and compensated for in its measurements. To produce surface plots similar to those in the previous section data were first split into flexor and extensor data. The data with angular velocities 5°/s or more below the intended velocity were removed; these were the data at the end of the range of motion when the direction of movement changed and the maximum velocity had not been reached yet. The eccentric data were given negative angular velocities and mean data were calculated for each angle-angular velocity combination. To have surface plots similar to those in the previous section data were linear interpolated for intermediate angular velocities.

The results of these measurements are presented below in surface plots of averaged data from three trials (Figure 4.23-Figure 4.28). The ankle plantar flexor torque measured on the dynamometer agreed to some extend with the values from literature (Figure 4.23), except for the higher torques at negative angular velocities. The eccentric joint torques measured at 180°/s were too high to be produced by the ankle plantar flexors and must therefore be measurement artefacts. It was therefore decided to leave the eccentric data measured at 180°/s out of the comparison. Production of peak eccentric isokinetic torque has been found difficult by others (Bellew & Yates, 2000). The ankle dorsi flexor torque (Figure 4.24) also showed high eccentric torque values at 180°/s and were also left out of the comparison. The torques for knee flexor and knee extensor torque measured with the dynamometer agreed well with those from literature (Figure 4.25 and Figure 4.26), except for the concentric knee extensor torque, where the measured values were lower than those found in literature. The torques for hip flexor and hip extensor torque measured with the dynamometer agreed well with those from literature (Figure 4.27 and Figure 4.28).

It can be concluded that the measured joint torque profiles for the ankle, knee and hip agreed well with the values based on data from literature, except for the concentric knee extensor torque and eccentric ankle torque at higher velocities. Eccentric ankle torque has been found previously to be difficult to measure; therefore the high values found for the highest angular velocity of eccentric ankle torque were considered to be measurement artefacts and not taken into the comparison. As there was a high variation in torque data found in literature the low values for concentric knee extensor torque were attributed to inter-individual variation. Pavol and Grabiner (2000) found high individual differences in knee torques measured on an isokinetic dynamometer, they also found knee starting angle and hip angle influenced the maximum torque measured at the knee. This high variability has to be kept in mind when using a single torque profile for different subjects and when comparing the torque profiles from literature with those measured on the younger female. It was concluded that the torque profiles based on data from literature agreed well enough with the measured data for a younger female, to use the torque profiles based on data from literature for both younger and older adults in the computer simulation model.

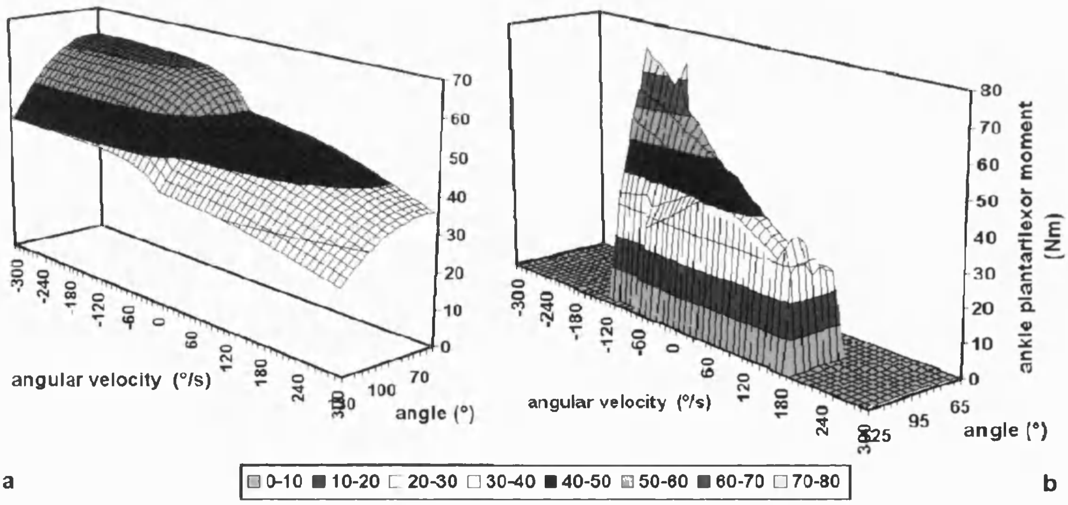


Figure 4.23 Surface plots of the torque-angle-angular velocity relationship for ankle plantar flexors, with a) values estimated for younger females based on data described by Gajdosik et al. (1999) and b) dynamometer data measured on a single young female.

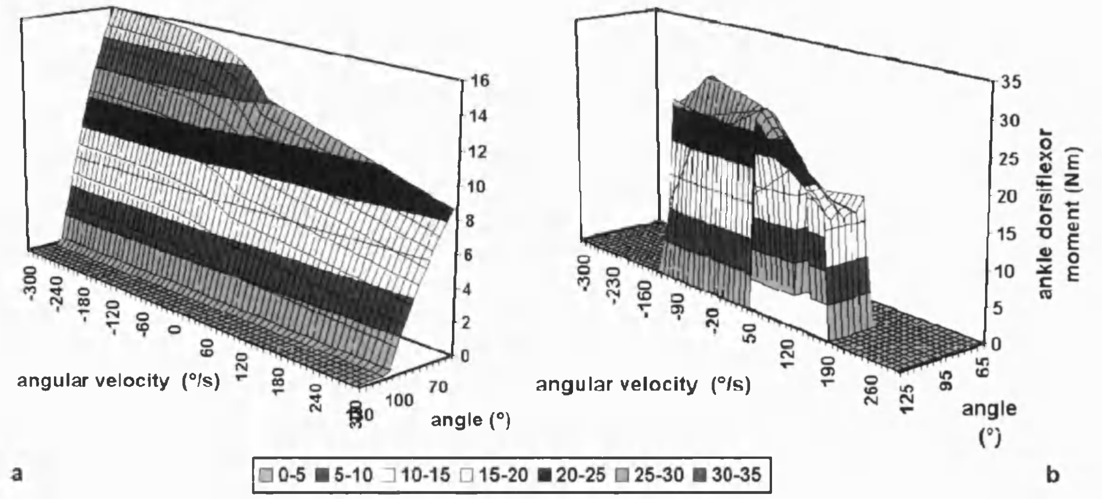


Figure 4.24 Surface plots of the torque-angle-angular velocity relationship for ankle dorsi flexors, with a) values estimated for younger females based on data by Wilson (2003) and b) dynamometer data measured on a single young female.

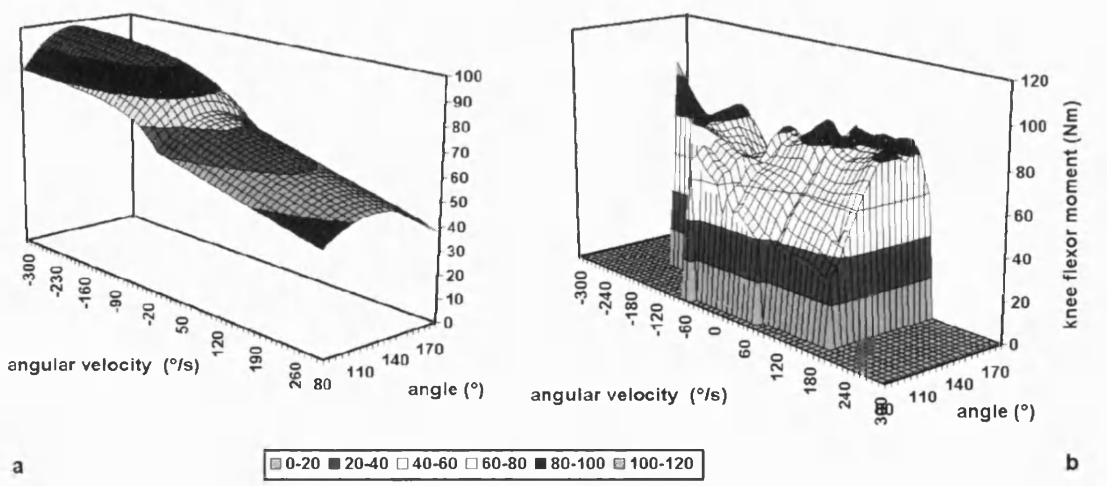


Figure 4.25 Surface plots of the torque-angle-angular velocity relationship for knee flexors, with on the left values estimated for younger females based on data by Wilson (2003) and Arnold et al. (Arnold et al., 1997) and on the right dynamometer data measured on a single young female.

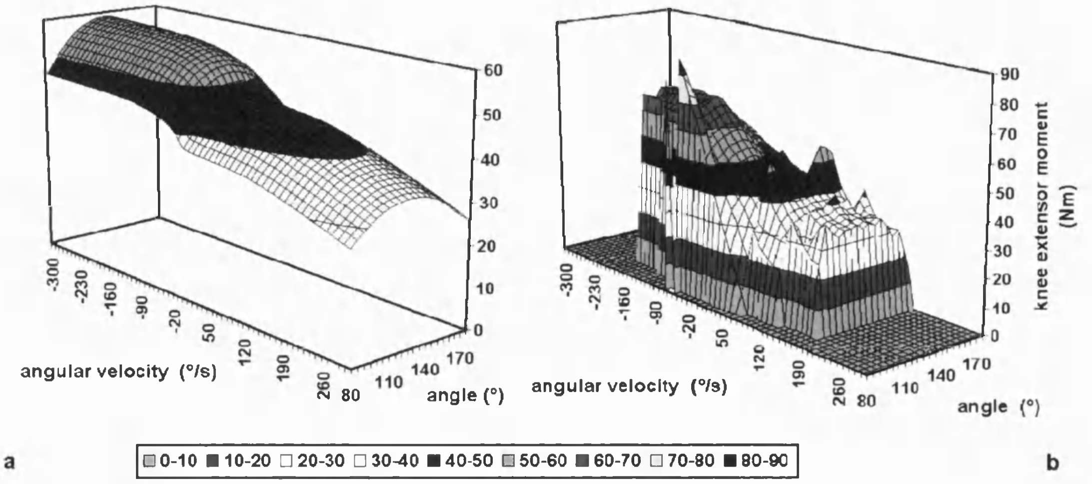


Figure 4.26 Surface plots of the torque-angle-angular velocity relationship for knee extensors, with a) values estimated for younger females based on values by Bellew and Yates (2000) and Wilson (2003) and b) dynamometer data measured on a single young female.

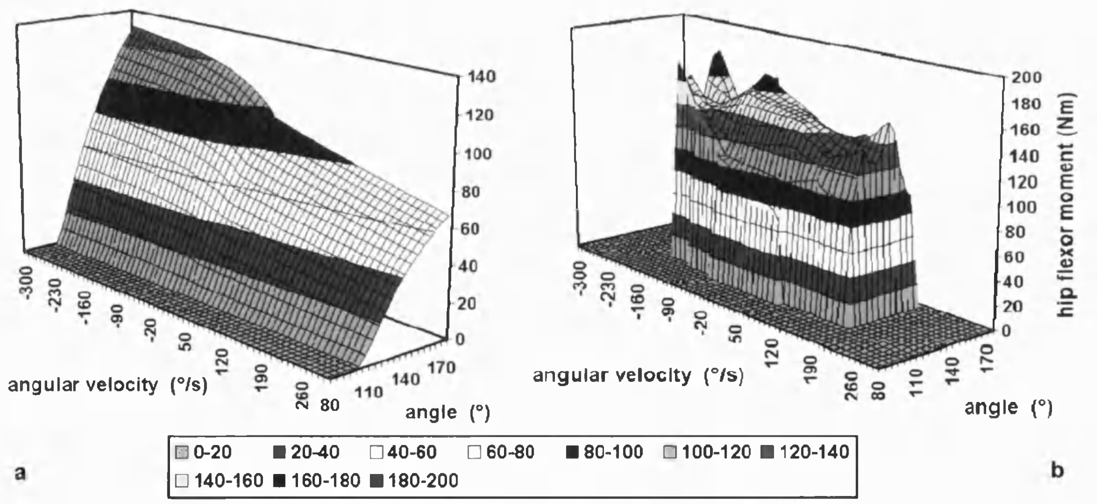


Figure 4.27 Surface plots of the torque-angle-angular velocity relationship for hip flexors, with a) values estimated for younger females based on values by Dean et al. (2004) and b) dynamometer data measured on a single young female.

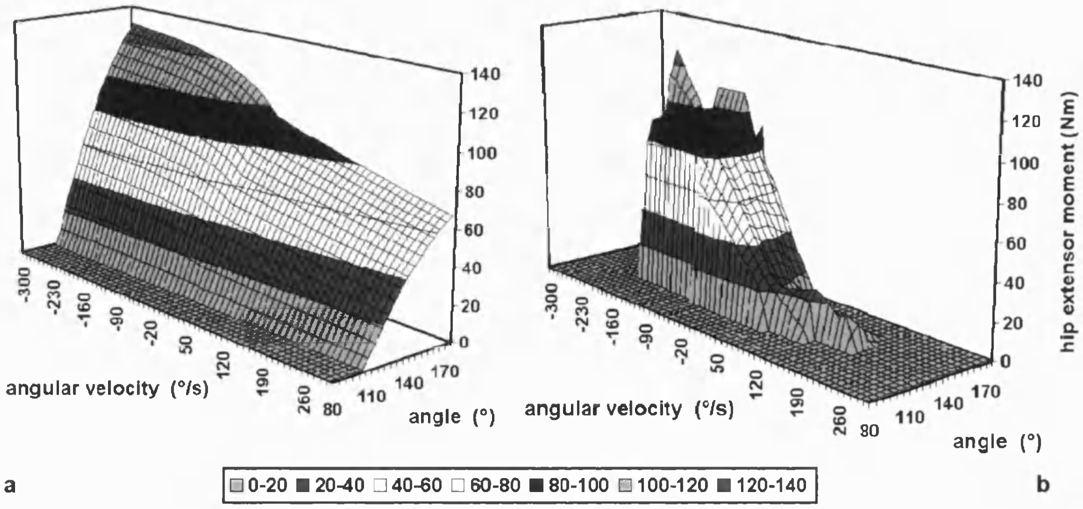


Figure 4.28 Surface plots of the torque-angle-angular velocity relationship for hip extensors of younger women, with a) values estimated for younger females based on values by Dean et al. (2004) and b) dynamometer data measured on a single young female.

4.6.4. Trip stimulus

A trip was induced in the model by applying a horizontal force (F_{trip}) to the toe of the swing foot. This force was applied for a certain length of time (t_{trip}) from time $t_{tripstart}$ and a function was developed to make the force profile F_{trip} resemble that measured in the trip recovery experiments. For this function the duration of the trip (t_{trip}), the maximum horizontal trip force ($F_{tripmax}$) and the time of this maximum force after the start of the trip ($t_{tripmax}$) were needed. The function started with a linear equation, making the force ramp up to the maximum trip force (Equation 4.12). After that the force returned to zero modelled with a reciprocal function with coefficient a_{trip} (Equation 4.13). Equation 4.14 made sure F_{trip} could not exceed the measured negative trip force. Coefficient a_{trip} was optimised with the Downhill Simplex method, minimising the RMS difference between the measured and simulated force, F_{trip}.

Equation 4.12 $$F_{trip} = \frac{(t - t_{trip}) * F_{trip\max}}{t_{trip\max}}$$ (when $t_{tripstart}<t<t_{tripmax}$)

Equation 4.13 $$F_{trip} = \frac{F_{trip\max} * t_{trip\max}}{a_{trip} * (t - t_{trip})}$$ (when $t_{tripmax}<t<t_{trip}-t_{tripstart}$)

Equation 4.14 $$F_{trip} = F_{trip\max}$$ (when $F_{trip}<F_{max}$)

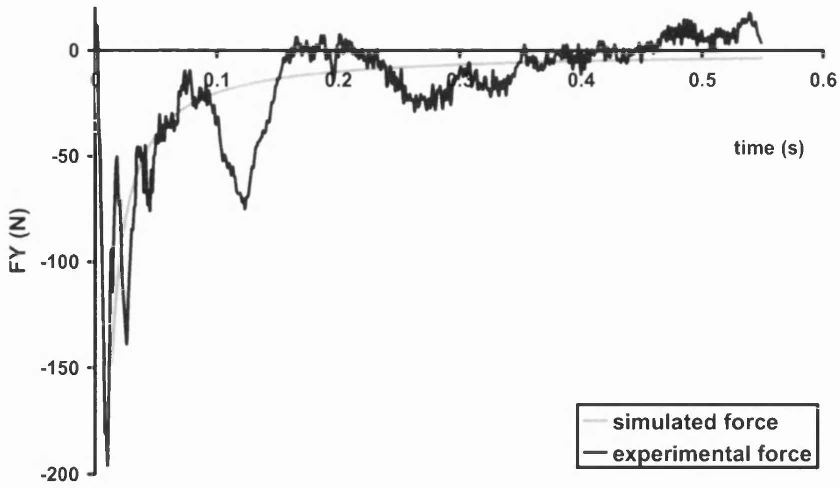


Figure 4.29 Simulated (grey solid line) and measured (black solid line) force during contact with the tripping device. The a_{trip} value for the simulated force was 0.59.

4.6.5. Muscle activation function

The torque profiles calculated with the nine parameter equation were maximum torque profiles. Torques produced during trip recovery and walking varied and were sub-maximal. To get more insight into what amount muscles were activated during trip recovery the EMG signals were investigated. It had to be kept in mind that EMG cannot be directly related to muscle force. For one younger subject EMG activity was calculated for each muscle relative to the overall maximum for that muscle, for walking, elevating strategy and lowering strategy recovery trials. This value gave an indication of the level of muscle activation. These values were averaged for each muscle group (ankle plantar flexors and dorsi flexors, knee flexors and extensors and hip flexors and extensors) and for three different phases of the walk or trip recovery (swing, impact phase of stance and push off phase of stance) (Figure 4.30, Figure 4.31 and Figure 4.32). Due to similar values across muscles but with high standard deviations no separate values were calculated for each muscle group and for each phase of walk and trip recovery. It was decided to calculate one value for all muscle groups for walking (a_{walk}) and another value for trip recovery (a_{trip}). The average EMG values only gave the ratio between a_{trip} and a_{walk}, which was 2.0. The nine parameter function was therefore multiplied by 0.5 prior to contact with the tripping device.

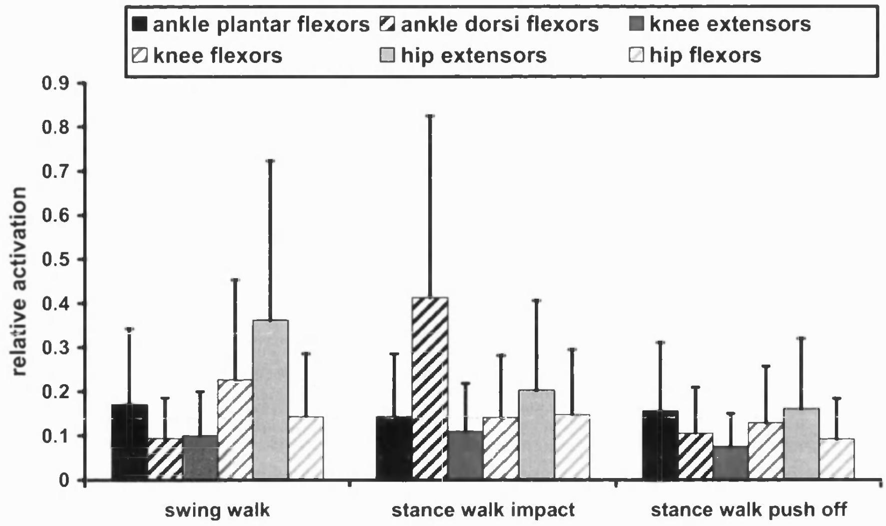


Figure 4.30 Relative muscle activation averaged for walking trials with standard deviation bars, for the swing phase, impact phase of stance and push off phase of stance. Values are given for ankle plantar flexors, ankle dorsi flexors, knee extensors, knee extensors, hip extensors and hip flexors.

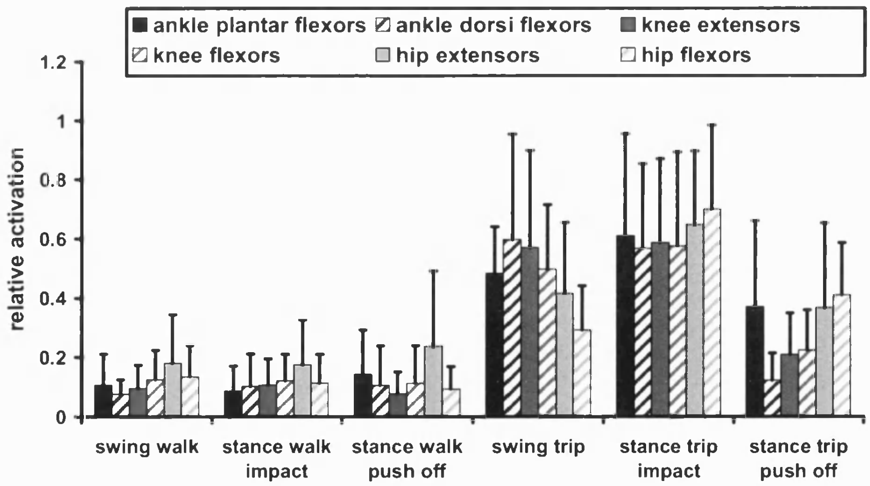


Figure 4.31 Relative muscle activation averaged for lowering strategy recovery trials with standard deviation bars, for the swing phase, impact phase of stance and push off phase of stance for both walk and trials. Values are given for ankle plantar flexors, ankle dorsi flexors, knee extensors, knee extensors, hip extensors and hip flexors.

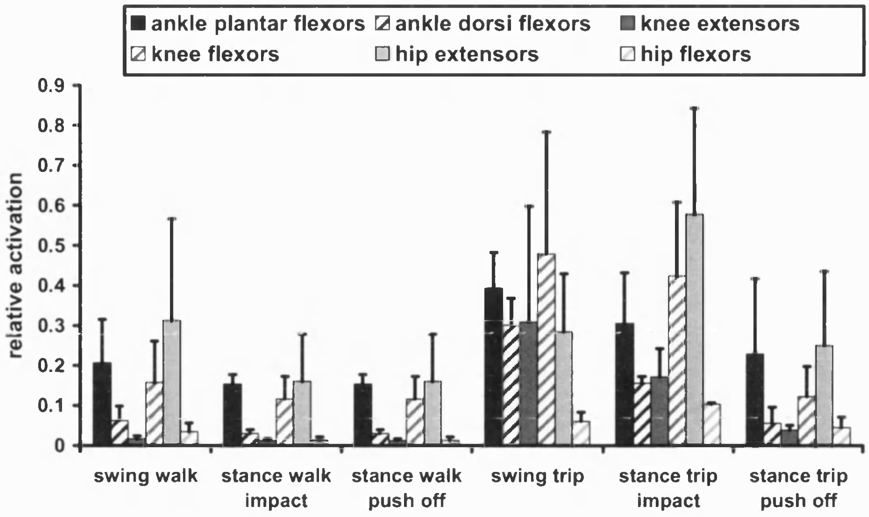


Figure 4.32 Relative muscle activation averaged for elevating strategy recovery trials with standard deviation bars, for the swing phase, impact phase of stance and push off phase of stance for both walk and trials. Values are given for ankle plantar flexors, ankle dorsi flexors, knee extensors, knee extensors, hip extensors and hip flexors.

The torque generators were not continuously activated during the simulations. An "on-off" function was created and the parameters for this function were optimised to match the model output with the experimental data in the model evaluation trials. While the activation function was "on" activation ramped up, using a quintic function, to a maximum activation. When the activation was "off" a second quintic function was used to make the activation ramp down to minimum activation. Based on the EMG data from the trip recovery experiment each torque generator was allowed to turn on and off twice during the simulations. Just after a trip stimulus activation was allowed to ramp up to a higher level (act4b). This was modelled using eight activation variables; act1: time when torque generator is initially switched on, act2: ramping time from minimum to maximum activation, act3: initial activation level, act4: the maximum activation level, act4b: the maximum activation level just after a trip, act5: time when torque generator is first switched off, act6: time when torque generator is switched on again, and act7: time when torque generator is finally turned off (Figure 4.33).

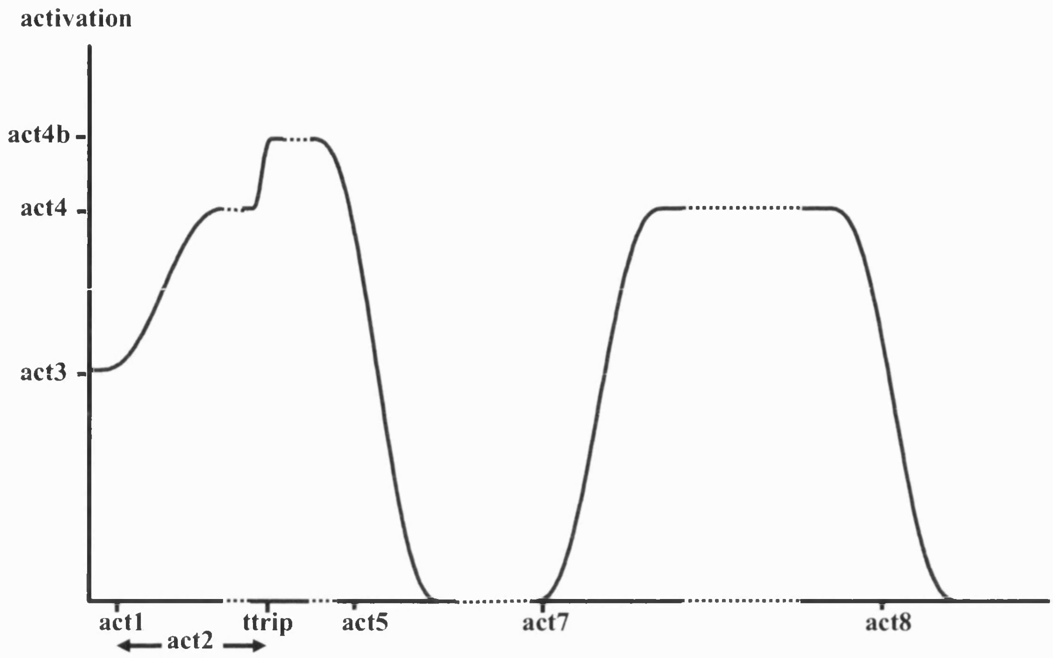


Figure 4.33 Schematic representation of the ramped activation function used to activate the joint torque generators.

4.7. Conclusions

Through several stages of development a ten-segment simulation model was developed to simulate trip recovery. The final model comprises horizontal and vertical spring-damper systems at the foot to simulate ground contact and flexor and extensor torque generators at the joints. These torque generators are actuated by a ramped activation function. Input data for the model were obtained experimentally (Chapter 3). To show the accuracy of the final model it will be evaluated and a sensitivity analysis will be performed to show sensitivity of the model to its main parameters (Chapter 5).

Chapter 5: Model evaluation and validation

5.1. Introduction

The trip recovery simulation model will be evaluated prior to any future simulations aiming to investigate the contributions to successful trip recovery. As stated in Chapter 2, model evaluation establishes quantitatively the level of accuracy that may be expected from a model (Yeadon & King, 2002).

In the model evaluation stage, variables were optimised such that the simulated and experimental results matched as closely as possible. Model evaluation and parameter optimisation were performed in four parts to reduce the number of variables in each optimisation; the spring-damper (section 4.6.1) and torque profile (section 4.6.2) parameters were optimised first, followed by model evaluation in two stages: prior to trip stimulus and during trip recovery. The second stage of the model evaluation is still ongoing. For each optimisation an objective function was defined that was minimised by the optimisation routine. These objective functions reflected the accuracy of the simulation and quantified how well simulated results matched experimental data.

5.2. Model evaluation

A total of 108 variables were optimised in the model evaluation; 96 to describe the ramped activation of the torque generators, three for initial vertical spring displacements, three for initial horizontal spring displacements and six for the initial angular velocities of the ankle, knee and hip joints.

All variables were optimised using the Simulated Annealing routine by Numerical Recipes (Press et al., 2002). This method was developed for continuous minimisation and the search area for the optimal solution was based on the Downhill Simplex method. It was chosen to use Simulated Annealing for this evaluation as this method is able to optimise problems with a large number of variables effectively; it is however computationally more expensive than the Downhill Simplex method (Press et al., 2002).

An objective function (Equation 5.2) was created that was minimised in the model evaluation. This weighted objective function rated the match between the simulated and experimental results and consisted of a number of components, including: the RMS difference of experimental and simulated joint angles time histories in ° (RMS_{angles}), a penalty if the joint angles exceeded the estimated RoM ($penalty_{ROM}$), the RMS difference of the experimental and simulated ground reaction force during the first recovery step as % (RMS_{GRF}), the RMS difference of the experimental and simulated CM position in m (RMS_{CM}) and a penalty if the simulation failed to converge ($penalty_{FC}$). This ensured body configuration was matched by the joint angles and body position by the body CM position.

As constraints to the optimisation, all variables were given upper and lower boundaries and the objective function was increased exponentially when variables were outside these boundaries (Equation 5.2). The upper and lower boundaries for the ramped activation parameters are shown in Table 5.1. The vertical spring displacements were allowed to vary 0.025 m around the initial position and the horizontal spring displacements 0.01 m, based on the maximum experimental spring displacements. The initial angular velocities were allowed to vary ±50°/s from the experimental values, as it was assumed these could not be derived from the experimental data as accurately as the initial angles and positions.

Equation 5.1 $penalty_{\exp} = -h_1 * e^{-10*h_1} - h_2 * e^{-10*h_2}$

where:

$h_1 = \text{var} - \text{var}_{\min}$

$h_2 = -\text{var} + \text{var}_{\max}$

var = optimised value of variable

$\text{var}_{\min}$ = minimum boundary value of variable

$\text{var}_{\max}$ = maximum boundary value of variable

The penalty that was given when a simulation failed to converge was relative to the time at which the simulation failed to converge, so a large penalty was given if the simulation failed to converge early in the simulation. The other components of the objective function were weighted as such that agreement of body position, joint angles, and ground reaction forces were accounted for in similar amounts in the objective function (Equation 5.2).

Equation 5.2

$$objective = 12 * RMS_{angles} + penalty_{ROM} + 2 * RMS_{GRF} + 6000 * RMS_{CM} + penalty_{FC} + penalty_{\exp}$$

Model evaluation was split into two parts to reduce the number of variables in a single evaluation; the first part evaluated the swing phase movement prior to the trip stimulus and the second part the trip recovery itself. The first part optimised the initial angular velocities, the spring displacements and some of the ramped activation variables, while the second part used these optimised variables and optimised the remaining ramped activation variables during trip-recovery. An $RMS_{angles} < 5.0°$ and an $RMS_{CM} < 0.01$ m were considered acceptable, based on the values accepted by King and Yeadon (2002; 2004; 2005; 2006). The model was evaluated with a representative trial of an elevating strategy recovery by an older adult.

Table 5.1 Upper and lower boundaries of the activation variables optimised in the model evaluation.

Variable	Lower boundary	Upper boundary
act1	-0.050	act5
act2	0.090	0.250
act3	0.000	0.500
act4	act3	1.500
act4b	act3	1.500
act5	act1	act7
act6	0.500	1.500
act7	act5	act8
act8	act7	1.200

5.2.1. Evaluation results

The first stage of the model evaluation was performed successfully, the second stage is ongoing. The components of the penalty function of the first stage of model evaluation are shown in Table 5.2. The RMS difference of the simulated and experimental joint angles

(RMS_{angles}) was 4°, which is below the maximum value of 5° that was deemed acceptable. The RMS difference of the CM positions (RMS_{CM}) was 0.006 m, which was also below the maximum value deemed acceptable (0.010 m).

Table 5.2 Components of the penalty function with their maximum value allowed in the evaluation and their value at the end of the first stage of the model evaluation.

Penalty	Maximum value	Evaluated value
RMS_{angles} (°)	5.0	4.0
RMS_{GRF} (%)	25	-
RMS_{CM} (m)	0.010	0.006
$penalty_{FC}$	0.0	0.0
$penalty_{ROM}$	0.0	0.0

The results of the first stage of the model evaluation are shown in Figure 5.1, with a stickfigure of the experimental data and the simulated data that were matched to these data in the model evaluation.

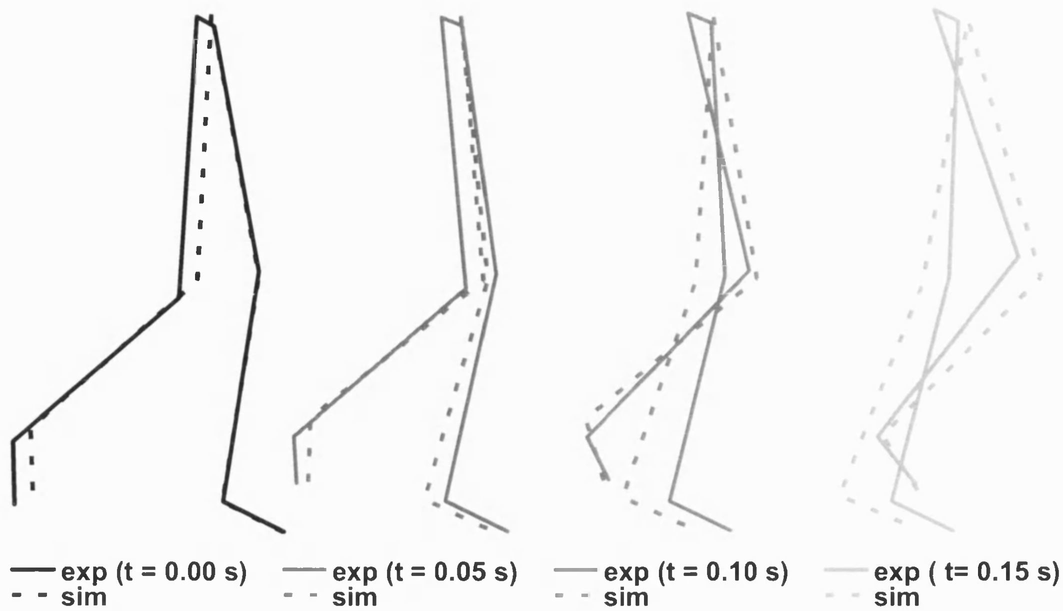


Figure 5.1 Stickfigure of experimental (solid lines) and simulated (dashed lines) data, resulting from the first phase of the model evaluation at 0.05 s intervals until the trip stimulus at t = 0.15 s.

5.2.2. Discussion

The results of the first phase of the model evaluation were within the limits deemed acceptable, and it can therefore be concluded that the trip recovery model was able to simulate the leg swing prior to the trip stimulus with sufficient accuracy. The second phase of the model evaluation is still ongoing. The trip stimulus appeared too abrupt to be processed by the contractile and elastic components of the torque generators in the swing ankle alone. The ankle joint angle changed so abruptly that the torque generators could not produce any joint moment. It is therefore suggested that passive torques are required before the joint exceeds its range of motion, as also at a neutral joint angle passive torques will be present (Gajdosik et al., 1996). The second phase of the model evaluation is ongoing and in the future will allow the simulation model to be applied to address additional research questions relating to trip recovery strategies. At present though, all of the research questions posed in this thesis could be addressed purely with the experimental data derived from the trip experiments.

Chapter 6: Contributions to successful trip recovery

6.1. Introduction

In this chapter the contributions to successful trip recovery determined from the experimental results are described. The principal research questions posed in Chapter 1 are addressed in separate sections and summarised at the end of this chapter with reference to the hypotheses.

6.2. Trial outcome measures

6.2.1. Outcome parameters

A trial was considered a failed trip recovery when over 30% of the body weight was supported by the safety harness. Of the total of 154 trip trials analysed 15 were classified as failed trip-recoveries. The occurrence of failed trip-recoveries was not significantly different between younger and older adults. The failed trip recovery trials had a significantly higher ($p<0.05$) average perturbation force than the successful recovery trials (206.3 N vs. 82.4 N).

The older adults had an average walking velocity of 1.11 LL/s, which was significantly lower ($p<0.05$) than that of the younger adults (1.22 LL/s). The older adults recovered significantly ($p<0.05$) more often with a lowering strategy than the younger adults. The older adults recovered in 72 out of 91 trip recovery trials with a lowering strategy and the younger adults in 26 out of 63 trials.

The primary outcome measure used to quantify how well someone recovered from a trip was the variable ”recovery amount”. This variable was used to quantify the reduction of the forward angular momentum around the centre of mass caused by a trip (i.e. frontal plane of the body is moving towards the floor). It was split into early and late recovery amount (ERA and LRA). ERA was defined as the difference between the maximum angular momentum between foot contact with the tripping device and ground contact of the recovery foot and the angular momentum at the instant of this ground contact. LRA was defined as the difference between the angular momentum at the instant of ground contact of the recovery foot and the minimum angular momentum during this ground contact. As positive angular momentum was defined in a forward direction, a larger recovery amount implied a larger reduction of the forward angular momentum that was caused by the trip. Also a separate recovery amount value for the arms was calculated; arm recovery amount (ARA). This value was calculated in the same manner as ERA, only for the angular momentum contribution of the arms instead of the whole body angular momentum.

No significant differences were detected in ERA between older and younger adults, or between elevating and lowering recovery strategies within the sub-groups (Table 6.1). This does not agree with findings by Pijnappels et al. (2005a) who found that during elevating strategy recoveries older adults were not able to restrain the forward angular momentum during the push-off. The positive average ERA of the older adults found in this thesis shows that most of the older adults were able to reduce their forward angular momentum. The younger adults showed a significantly ($p<0.05$) higher LRA in elevating than in lowering strategy recoveries (0.0011 and 0.006 m/s respectively), while older adults showed a significantly ($p<0.05$) higher LRA in lowering than in elevating strategy recoveries (0.009 and 0.004 m/s respectively). Only LRA of the elevating strategy recoveries was significantly ($p<0.05$) different between younger and older adults (0.011 and 0.004 m/s respectively).

*Table 6.1 Average ERA and LRA values (normalised units) for younger and older adults with standard deviations. Significant differences to younger subjects ($p<0.05$) are indicated with *, and to elevating strategy recovery trials with $^{\&}$.*

		ERA (m/s)	LRA (m/s)
Younger	Elevating	0.004±0.004	0.011±0.007
	Lowering	0.004±0.004	$0.006\pm0.005^{\&}$
Older	Elevating	0.005±0.003	0.004±0.002*
	Lowering	0.004±0.006	$0.009\pm0.004^{\&}$

6.2.2. Discussion

The slower walking velocity found for older adults agrees with findings in literature (Payne & Isaacs, 1987; Spirduso, 1995; Shephard, 1997). The finding that the older adults more often adopted a lowering strategy than the younger adults is also in agreement with findings by Pijnappels et al. (2005a). The underlying reasons to why older adults prefer a lowering strategy recovery have not been described in literature. Some possible underlying reasons were found in the present study and will be discussed in Chapter 7.

The younger and the older adults had similar ERA values, which means they reduced their forward angular momentum by similar amounts during contact of the initial stance limb. This is not in agreement with Pijnappels et al. (2005a), who showed that older adults reduced angular momentum insufficiently during contact of the initial stance limb, which would mean older adults should have a smaller ERA than younger adults. It is unclear what would cause this discrepancy, as the experimental protocols of both studies were similar. The participants in this study were slightly older than those in the non-fallers participant group of Pijnappels et al. (2005a) (70.0 vs. 66.5 years) and they were all female in this study while Pijnappels et al. used mainly male participants. It can however be expected that the difference in reduction of angular momentum would be more present in an older participant group. A gender effect could be an underlying cause of the difference found, although gender differences in the biomechanics of trip recovery have not been described fully in the literature. The number of older participants was small in both studies (four in Pijnappels et al.'s study and seven in this thesis) and trip recovery movements have been found to have large variation, so a larger participant group might clarify the differences found.

6.3. Joint moments

This section addresses the first research question "What is the contribution of the recovery limb in successful trip recovery in both younger and older adults?". In doing so the joint moments of the recovery limb during trip recovery and walking were investigated.

6.3.1. Experimental results

A redistribution of joint moments with increasing age has been found during normal gait (Savelberg et al., 2007). To get a better idea of the joint moment distribution during trip recovery, the support moment impulse was calculated as the sum of the integral of the moments at all three joints, with extensor moments as positive (Winter, 1990). The integrated moments at the ankle, knee and hip joint were compared with each other to investigate whether a redistribution of joint torques was present. Ankle, knee and hip joint moments of the recovery limb during elevating and lowering strategy recoveries and of the stance limb during walking of both younger and older adults from representative trials are shown in Figure 6.1.

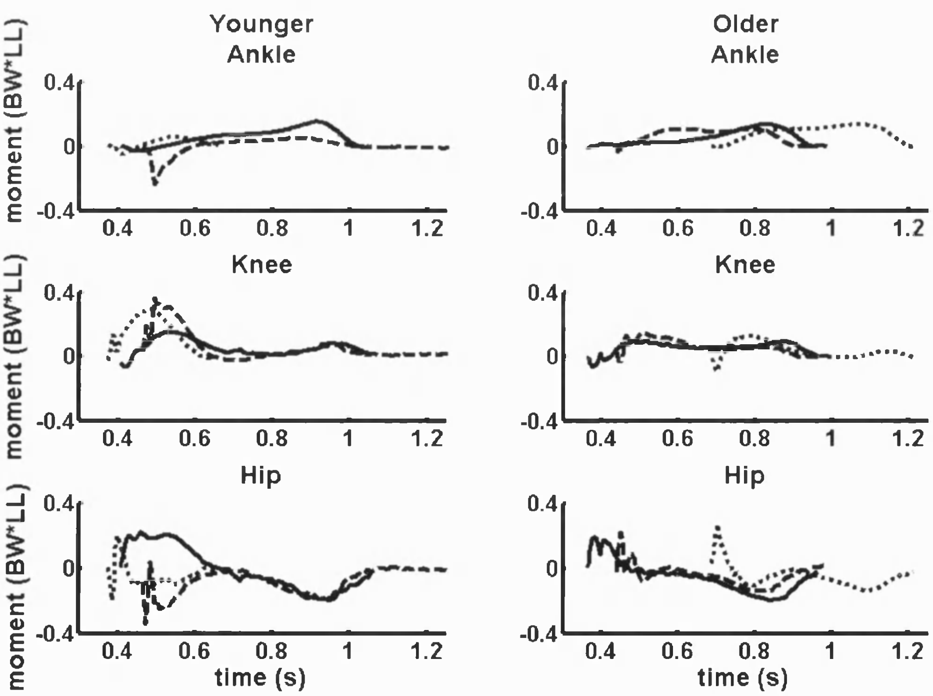


Figure 6.1 Normalised recovery limb joint moments of representative trials from younger and older adults. The time scale is such that contact with the trip-device occurs at 0.0 s.

No significant differences were found between younger and older adults for the average peak joint moments at any joint (Table 6.2). Both the younger and the older adults showed significantly larger ($p<0.05$) peak moments at the knee during lowering strategy recoveries than during walking (Table 6.2). The younger adults showed significantly larger ($p<0.05$) moments at the hip during elevating strategy recoveries than during walking (Table 6.2). The peak moment of the hip of the recovery limb was significantly different ($p<0.05$) for lowering strategy trials than for elevating strategies for the younger adults (Table 6.2). The peak hip joint moment was an extensor moment during elevating strategy recoveries, which would maintain the upper body upright, and a flexor moment during lowering strategy recoveries, which possibly served to further lower the body CM. The older adults did not show significantly different maximum joint moments during elevating and lowering strategies. No correlation was found between the maximum joint moments and LRA for both the younger and the older adults.

*Table 6.2 Peak moments (relative to BW*LL) for the joints of the recovery limb for younger and older subjects. Trials are walk trials and elevating and lowering recovery strategy trials. Significant differences to younger subjects ($p<0.05$) are indicated with *, to walk trials with $^{+}$, and to elevating strategy recovery trials with $^{\&}$.*

		Maximum moment (BW*LL)		
		Ankle	Knee	Hip
Younger	Walk	0.17± 0.28	0.10± 0.13	-0.03± 0.19
	Elevating	0.12± 0.04	0.21± 0.15	$0.33\pm 0.13^{+}$
	Lowering	0.19± 0.38	$0.23\pm 0.19^{+}$	$-0.05\pm 0.36^{\&}$
Older	Walk	0.19± 0.11	0.04± 0.15	0.06± 0.20
	Elevating	0.04± 0.17	0.15± 0.05	0.14± 0.23
	Lowering	0.21± 0.25	$0.20\pm 0.11^{+}$	0.10± 0.29

The younger adults showed a higher total support moment impulse during elevating strategies than during walking ($p<0.05$) while the older adults did not, but no significant differences were found for the total support moment impulse between younger and older adults in walking, elevating and lowering strategy recoveries (Figure 6.2). The older adults showed a higher knee moment impulse during lowering than during elevating strategy recoveries ($p<0.05$), while the support moments of the individual joints in the younger adults did not differ significantly between the different strategies (Figure 6.2).

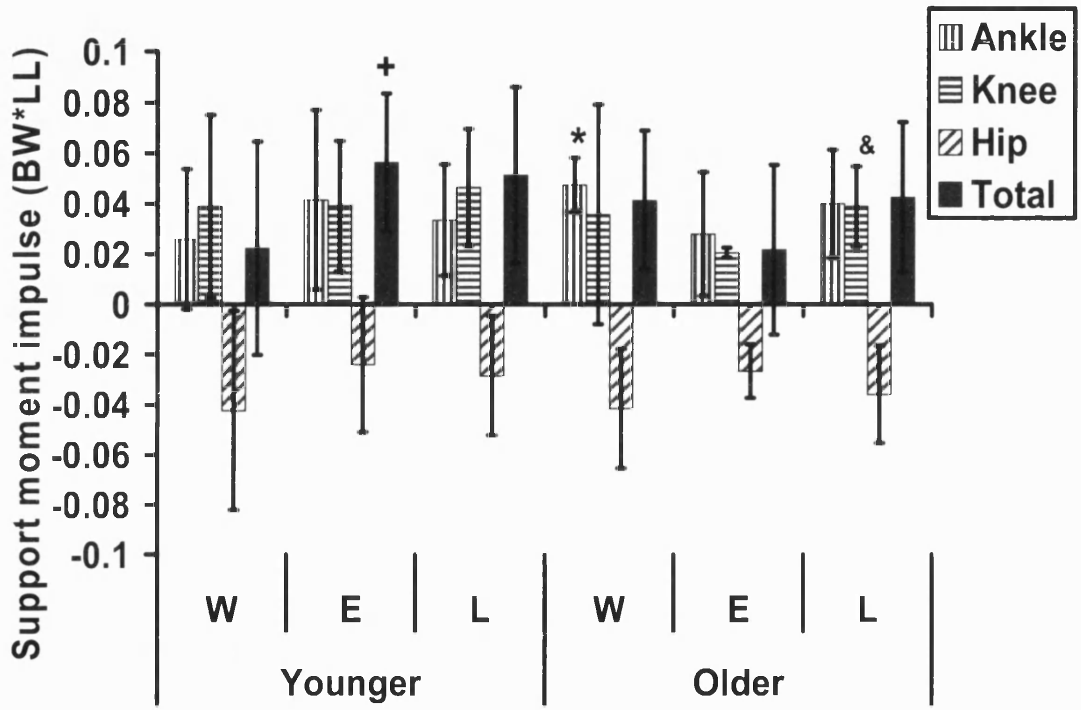


*Figure 6.2 Moment impulse for the individual joints and total support moment impulse for younger and older adults, for walk (W), elevating (E) and lowering (L) strategy recovery trials. Significant differences to younger subjects ($p<0.05$) are indicated with *, to walk trials with + and to elevating strategy trials with $^{\&}$.*

6.3.2. Discussion

Peak joint moments were not correlated with LRA. This means a higher peak joint moment did not necessaryly lead to a better trip recovery. During a trip a relatively higher moment was produced at the knee of the recovery limb than at the ankle.

The younger adults had an increased support moment during elevating strategy recoveries relative to walking, while the older adults did not. LRA was during elevating strategy recoveries smaller in older adults than in younger adults. This suggests the older adults could not create sufficient support moment to reduce their forward angular momentum. The older adults showed a significantly higher knee moment impulse during lowering than during elevating strategy recoveries, while the younger adults showed similar knee moment impulses during elevating and lowering strategy recoveries. This indicates the older adults were not able to produce a sufficiently high support moment at the knee during elevating strategy recoveries and therefore could be one reason why they more often adopt a lowering strategy. The inability to produce a sufficiently high support moment at the knee could be due to placement of the recovery limb at ground contact; a slower response time and movement velocity may result in an inability to place the recovery limb to allow production of sufficient support moment at the knee to recover from a trip.

The total support moments were similar during all conditions in the older adults, although the ankle, knee and hip joints contributed in different amounts. During both elevating and lowering strategy recoveries in both younger and older adults the majority of the support moment was produced by the ankle and knee.

The older adults produced similar peak joint moments during trip recovery to the younger adults, the younger adults however recovered better. This raises the question as to whether peak muscle strength is a limiting factor in trip recovery. Would it be better to focus on the development of joint power or on recovery technique? It has already been proposed by Skelton et al. (2002) that muscle power might be a better predictor for fall risk than muscle strength. The younger adults produced a higher support moment during trip recovery than during walking, which the older adults did not, and the separate joints contributed differently to the support moment in younger and older adults. This leads to the suggestion that a change of recovery technique might lower the risk of falling.

6.4. Muscle activation and sequencing

This section addresses the second research question "How do muscle sequencing and coactivation influence successful trip recovery in both younger and older adults?". It investigates and describes the EMG signals measured in the trip recovery experiment.

6.4.1. Experimental results

In the trip recovery experiment all EMG electrodes were placed on one limb only. Therefore in each trial EMG data were collected either from the initial stance or the recovery limb. A rectified raw EMG signal is shown in Appendix E.

Muscle sequencing was investigated by comparing the RMS EMG signals relative to the average activity during walking. The signals were time-normalised in such a way that 0% was at contact with the tripping device and 100% at contact of the recovery foot with the floor. Figure 6.3 and Figure 6.4 show the mean of these resampled signals for all analysed trials together with plus or minus one standard deviation curves. The standard deviations of the RMS signals were in most muscles smaller in the older than in the younger adults, except for TA (Figure 6.3 and Figure 6.4). This suggests that the older adults had a more consistent response, which was similar for most older adults and in most trials (except for tibialis anterior), while the younger adults showed varying responses. The muscle sequencing will be described refered to Figure 6.3 and Figure 6.4. During elevating strategies both younger and older adults activated TA, which is a dorsiflexor. This was probably to provide toe-clearance from the obstacle and to counteract the plantarflexor movement caused by contact with the tripping device. The older adults also activated TA during ground contact, probably to increase ankle stiffness as the plantarflexors (GA and SO) were also activated during ground contact. The younger adults activated the plantarflexors (GA and SO) only during ground contact. This was assumed to be to reduce the forward angular momentum of the body and to create a push off moment at the end of ground contact. For elevating strategy recoveries, the younger adults activated SM and BF just after contact with the tripping device. It was assumed that they were acting here mainly as knee flexors, which would lift the recovery foot over the obstacle. Both the younger and older adults activated SM and BF (knee flexors and hip extensors) during ground contact, presumably acting as hip extensors here to maintain the trunk upright. VL (knee extensor) was not much activated in both younger and older adults. The younger adults activated RF (knee extensor and hip flexors) just after contact with the tripping device (for obstacle avoidance), whereas the older adults activated RF at the end of ground contact, to generate a push off reaction. The signal of GM was in most trials not good and will therefore be left out of the description of muscle sequencing.

During lowering strategy recoveries both younger and older adults activated TA (dorsiflexor) during ground contact. As the plantarflexors (GA and SO) were also activated during ground contact it can be suggested that there was muscle coactivation to increase stiffness at the ankle joint. The older adults also activated TA just after contact with the tripping device, which can be expected to dorsiflex the foot and increase obstacle clearance. Both the younger and the older adults activated GA and SO (plantarflexors) prior to and during ground contact, this would serve to orient the foot correctly for ground contact since the foot was initially dorsiflexed for obstacle clearance. GA and SO were active during ground contact, this was probably to reduce the forward angular momentum of the body and to create a push off reaction at the end of contact. The younger adults activated SM and BF (knee flexors and hip extensors) prior to and during ground contact, while the older adults activated them during ground contact only. During ground contact these muscles would act as hip extensors to maintain the trunk in an upright position. Both the younger and the older adults activated VL just after contact with the tripping device and during ground contact. They both activated RF during ground contact, the younger adults also activated RF just prior to ground contact. VL and RF are knee extensors and RF is also a hip flexor. RF was expected to act just after contact with the tripping device as hip flexor, to lift the limb over the obstacle, and during ground contact both VL and RF were expected to act as knee extensors to absorb impact and maintain the body in an upright position.

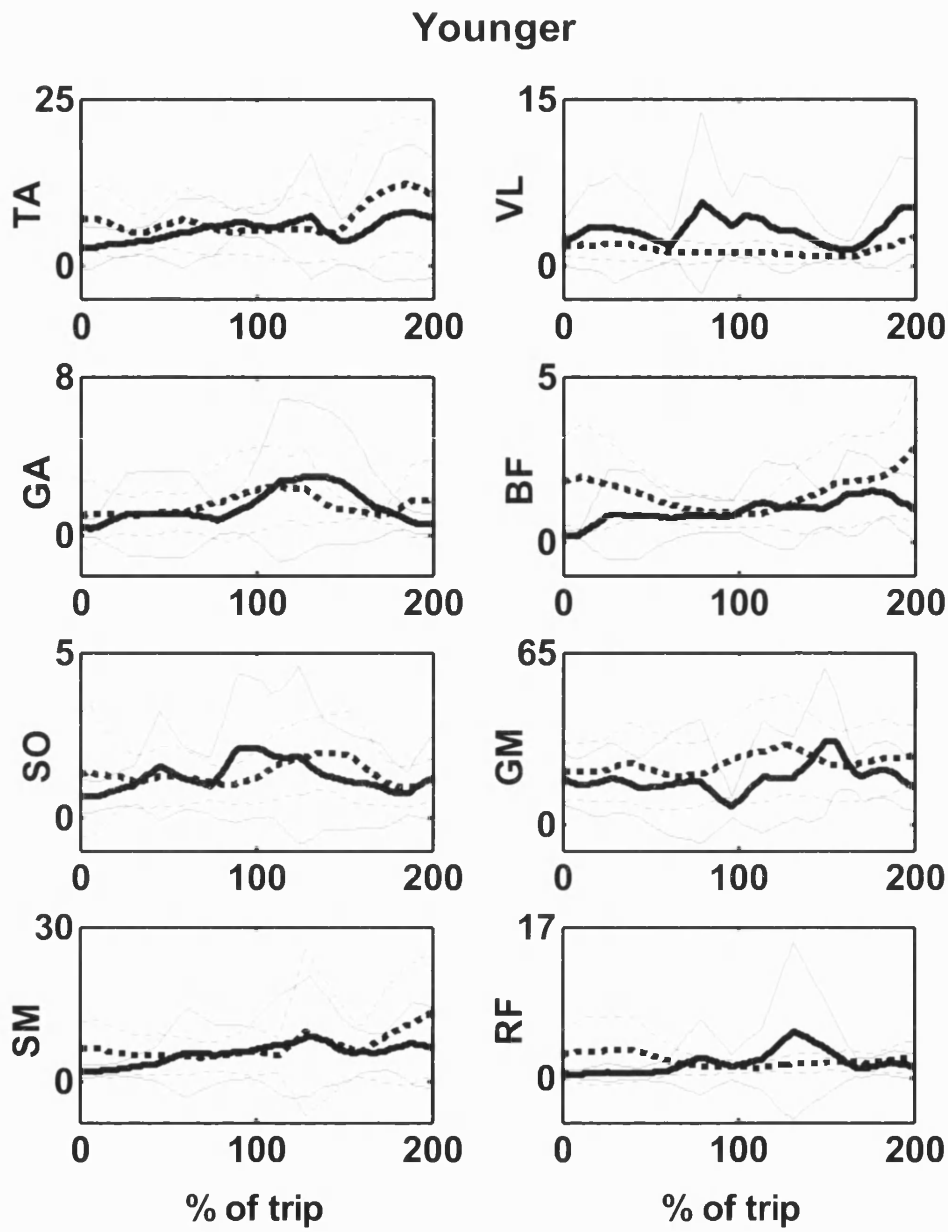


Figure 6.3 RMS EMG data for younger adults. The dotted lines are for elevating strategy recoveries and the solid lines for lowering strategy recoveries. The wide lines are the mean values and the narrow lines are the mean plus or minus one standard deviation. TA is the tibialis anterior, VL the vastus lateralis, GA the gastrocnemius, BF the biceps femoris, SO the soleus, GM the gluteus maximus, SM the semimembranosus and RF the rectus femoris. The data are time rectified, such that 0% of trip is at contact with the tripping device and 100% is at ground contact of the foot of the recovery limb.

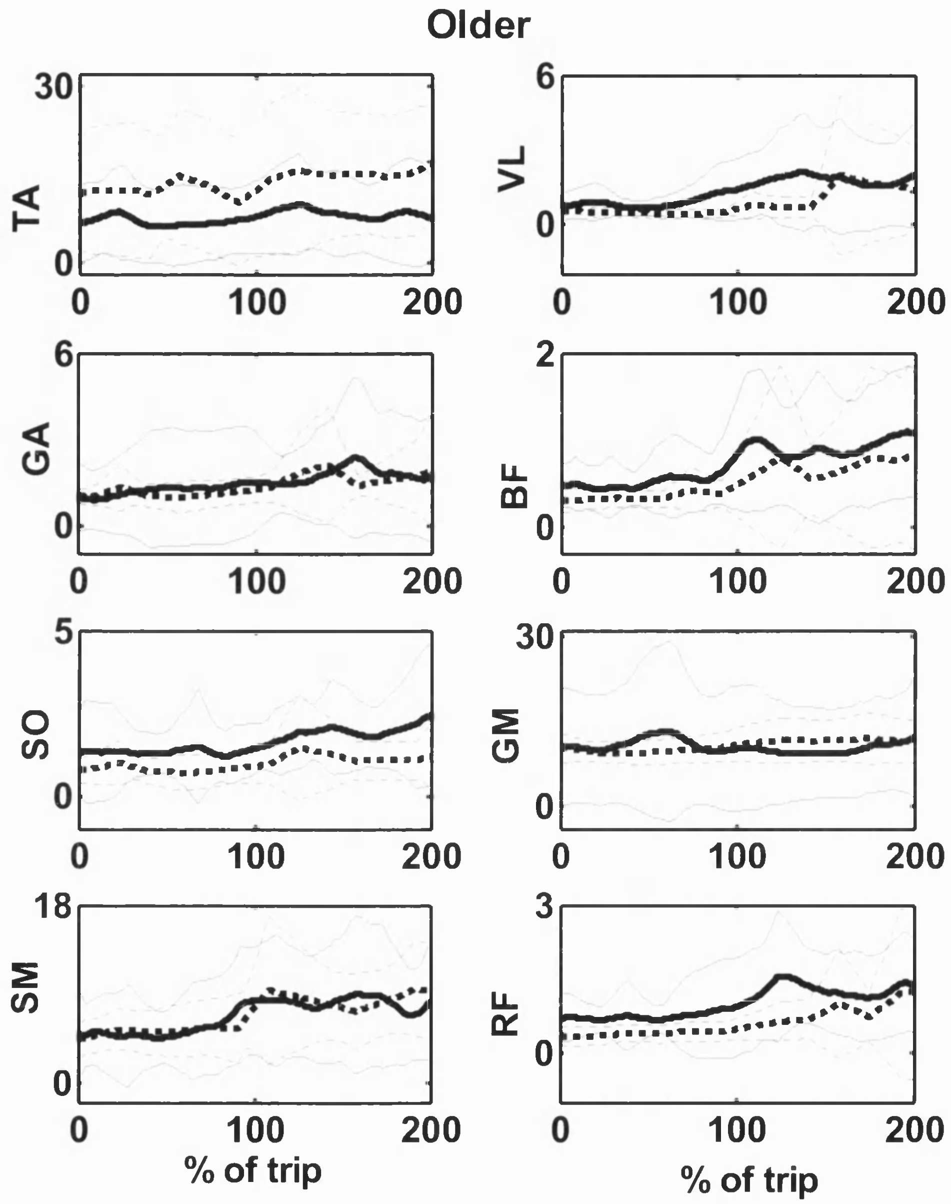


Figure 6.4 RMS EMG data for older adults. The dotted lines are for elevating strategy recoveries and the solid lines for lowering strategy recoveries. The wide lines are the mean values and the narrow lines are the mean plus or minus one standard deviation. TA is the tibialis anterior, VL the vastus lateralis, GA the gastrocnemius, BF the biceps femoris, SO the soleus, GM the gluteus maximus, SM the semimembranosus and RF the rectus femoris. The data are time rectified, such that 0% of trip is at contact with the tripping device and 100% is at ground contact of the foot of the recovery limb.

Muscle onset timings of the lower limb muscles were similar for the younger and older adults during elevating and lowering strategy recoveries. Standard deviations of the muscle onset timings were high for younger and older adults for both elevating and lowering strategy recoveries (Figure 6.5). This means there was a large variation in muscle onset timings. During elevating strategy recoveries the older adults showed a significantly ($p<0.05$) earlier onset time than the younger adults in the SM of their recovery limb (Figure 6.5). This muscle extends the hip and flexes the knee; an earlier onset time should result in a better placement of the recovery limb to assist trip recovery. This better placement of the recovery limb due to an ealier onset time may in older adults however be limited by movement speed. Muscle onset timings were similar for elevating and lowering strategy recoveries (Figure 6.5). The younger adults showed however a significantly ($p<0.05$) faster response in the GA of their recovery limb during elevating than during lowering strategy recoveries (Figure 6.5). This muscle plantarflexes the ankle, as in elevating strategy recoveries the recovery foot contacts the ground sooner after contact with the tripping device than in lowering strategy recoveries it can be expected that the plantar flexors needed to be activated earlier to reduce the forward angular momentum of the body. Older adults showed a significantly slower response in the BF of their recovery limb during elevating than during lowering strategy recoveries (Figure 6.5). This muscle flexes the knee and extends the hip, so a relatively earlier onset time in younger adults may result in a better ability to keep the trunk upright.

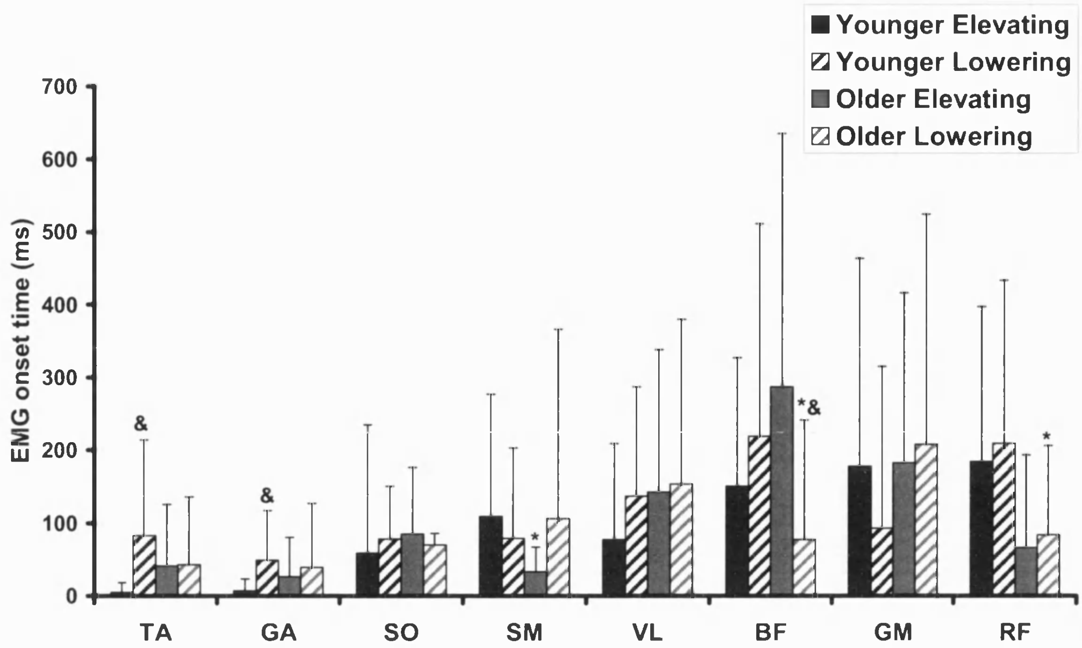


*Figure 6.5 Average EMG onset timing after trip stimulus for tibialis anterior (TA), gastrocnemius (GA), soleus (SO), Semimembranosis (SM), vastus lateralis (VL), biceps femoris (BF), gluteus maximus (GM) and rectus femoris (RF) of the recovery limb for younger and older adults and elevating and lowering strategy recoveries. Significant differences ($p<0.05$) to younger subjects are indicated with *, and significant differences to elevating strategy recovery trials with* [&].

Recovery step time (time between contact with tripping device and contact with force plate) was not significantly different between younger and older adults ($p<0.05$). Recovery step time was found to be significantly lower ($p<0.05$) for elevating (younger: 0.40±0.17 s, older: 0.34±0.20 s) than for lowering strategy recoveries (younger: 0.56±0.26 s, older: 0.49±0.16 s) for both younger and older adults. EMG onset timing was not correlated with the response time in the response test, or foot sensation. This means the older adults who responded faster in the test and had lower sensory loss did not activate their muscles earlier during trip recovery. EMG onset timings were also not correlated with ERA and LRA, so earlier muscle onset timing did not result in a larger reduction of the forward angular momentum.

Coactivation at the ankle, knee and hip of the recovery limb around impact are shown in (Table 6.3). In both the younger and the older adults coactivation was mainly present at the ankle and hip and was small at the knee, during walking and during trip recovery. In the younger adults coactivation during trip recovery did not differ significanty ($p<0.05$) from that during walking. The older adults showed a significantly ($p<0.05$) lower coactivation at the ankle during lowering strategy recoveries than during walking and elevating strategy recoveries, and a significantly ($p<0.05$) higher coactivation at the hip during elevating and lowering strategy recoveries than during walking. The older adults showed a significantly ($p<0.05$) higher coactivation than the younger adults at the ankle during walking and during elevating strategy recoveries.

*Table 6.3 Average coactivation values for the ankle, knee and hip of the recovery leg. Significant differences ($p<0.05$) to younger subjects are indicated with *, to walk trials with $^+$, and to elevating strategy recoveries with $^{\&}$.*

		Recovery leg		
		$coact_{ankle}$	$coact_{knee}$	$coact_{hip}$
Younger	Walk	3.6±3.3	1.0±2.3	10.7±12.3
	Elevating	3.6±3.2	0.4±0.7	12.7±9.9
	Lowering	4.4±4.2	0.8±1.1	12.2±15.8
Older	Walk	14.9±10.0*	2.1±4.9	5.7±7.4
	Elevating	20.2±11.0*	0.6±0.2	$11.0±3.5^{+}$
	Lowering	$7.5±6.6^{\&+}$	2.2±3.7	$11.9±10.6^{+}$

The correlation of LRA with the coactivation was investigated, because LRA was calculated during contact of the recovery foot with the ground. It was expected that, providing the recovery step is large, a higher coactivation, especially at the ankle joint, would increase the reduction of the forward angular momentum of the body. Coactivation was not correlated with LRA for the older adults.

6.4.2. Discussion

Muscle sequencing differed between the younger and the older adults for elevating and lowering strategy recoveries. In elevating strategy recoveries both younger and older adults activated ankle dorsiflexors after contact with the tripping device, probably to provide toe-clearance from the obstacle. The older adults appeared to attempt to stiffen their ankle during ground contact as both plantar- and dorsiflexors were activated (Hortobagyi & DeVita, 2000). As the data presented in Figure 6.3 and Figure 6.4 are averaged data and the standard deviations were high it can be assumed that muscle sequencing varied a lot and activation of both flexors and extensors in the average data can only suggest that there might be coactivation. These data were therefore not compared with the coactivation data in Table 6.3, which are the averages of coactivation of individual trials. The younger adults appeared to attempt to reduce the forward angular momentum of their body during elevating strategy recoveries by activating their ankle plantarflexors only, and not by coactivation of plantar- and dorsiflexors. The younger adults activated their knee flexors just after contact with the tripping device, probably to provide toe-obstacle clearance, which the older adults did not. Both younger and older adults activated their hip extensors during ground contact to maintain the trunk upright. Hip flexors were activated in both younger and older subjects to swing the obstructed limb over the obstacle. The older adults activated their knee extensors at the end of ground contact, which the younger adults did not.

During lowering strategy recoveries, the older adults activated their dorsiflexors after contact with the tripping device, which can be expected to provide them extra toe-clearance with the obstacle. Both younger and older adults activated their plantarflexors prior to and during contact, in order to position the foot prior to contact and to reduce the forward angular momentum of the body during contact. Hip extensors were activated during contact to maintain the trunk upright, the younger adults also activated them prior to contact, which indicates they prepared for contact. Hip flexors were, in both younger and older adults, activated just after contact with the tripping device to swing the recovery limb over the obstacle. Knee extensors were activated during ground contact, and in the younger adults also prior to ground contact, probably to maintain the body in an upright position by extending the leg and to absorb impact.

It was suggested that the older adults had more consistent muscle activation responses within elevating and within lowering strategy recoveries, which was similar for most older adults and in most trials of the same recovery strategy (except for TA), while the younger adults showed varying responses. The younger adults used different options in trip recovery and may therefore be more adaptable, given that there is variation within the recovery strategies due to different situations and input conditions.

Response time in general decreases with age (Payne & Isaacs, 1987; Spirduso, 1995; Shephard, 1997). Lockhart et al. (2005) found an overall slower recovery process in older than in younger subjects when recovering from a slip. While Thelen et al. (2000) found no difference in muscle onset timing between younger and older adults and Pijnappels et al. (2005b) only found a significant difference between younger and older adults for onset timing for SO and not for GM, BF, RF, VL, TA and GA. They proposed prevention of falls in older adults to be affected by both lower extremity muscle strength and sensory degradation, rather than by major changes to muscle sequencing.

The muscle onset timings found in this study were similar between the younger and the older adults for both elevating and lowering strategy recoveries. This agrees with findings by Pijnappels et al. (2005b) and Thelen et al. (2000). Pijnappels et al. (2005b) found muscle latency times of 60-130 ms during trip recovery and Thelen et al. (2000) found muscle latency times ranging from 73-114 ms during stepping responses to regain balance, most of the muscle onset timings found in this study were in the same range, some were however smaller. This is probably because these muscles were sometimes already activated at the time of the trip stimulus, bringing down the average onset time. Earlier onset times were found for the older than the younger adults for the SM of their recovery limb in elevating strategy recoveries and the BF and RF of their recovery limb during lowering strategy recoveries. It was found in section 6.3 that, during elevating strategy recoveries, the support moment in the older adults was limited by the moment at the knee. The early activation of BF and RF is possibly to attempt to create a higher stiffness at the knee.

Recovery step time (time between contact with tripping device and contact with force plate) was not significantly different between younger and older adults (younger elevating: 0.40 s, younger lowering: 0.56 s, older elevating: 0.34 s and older lowering: 0.49 s). These times were shorter, but similar, than the time found by Pavol et al. (2001) for older subjects (0.45 s for elevating and 0.52 s for lowering strategy recoveries). While recovery step time was similar for younger and older adults, younger adults had a larger recovery step than older adults (section 6.7). This leads to the suggestion that movement velocity is more important in trip recovery than response time. Recovery step time was found to be significantly lower ($p<0.05$) for elevating than for lowering strategy recoveries for both younger and older adults. With the definition used in this thesis a longer step time was expected for lowering strategy recoveries than for elevating strategy recoveries as the obstructed limb first needs to be lowered to the ground.

It was expected that older adults who responded faster in the test and had lower sensory loss would activate their muscles earlier during trip recovery and that this would result in a larger reduction of the forward angular momentum. However EMG onset timing was not correlated with the response time in the response test time, foot sensation, ERA or LRA. This might be as the older adults in this study were all healthy and a slowed response time and sensory loss might not have been present in such an amount that muscle onset timing would be affected. Foot sensation may not have been correlated with recovery amount because contact with the trip device is far above the minimum threshold for foot sensation. The foot sensation threshold is more important for more subtle adjustments. It also has to be kept in mind that EMG amplitude is not directly correlated with muscle force; a muscle can have an early onset time but still produce little force.

Hortobágyi and de Vita (2000) investigated leg stiffness and ageing in downward stepping movements. They found that older adults increased their leg stiffness by coactivation of BF and TA. Increasing muscle coactivation will increase energy expenditure, and it has been suggested that coactivation may serve to optimise stiffness (Hasan, 1985). Both the younger and older adults showed relatively high coactivation at the ankles and hips and not at the knees. Minimal coactivation was found at the knee as a high extensor moment would be required to keep the body upright. The moment at the knee was however found to be relatively small in older adults compared with that in younger adults (section 6.3); this might be due to limited maximum muscle force in older adults or a placement of the recovery limb in which moment production was restricted. Coactivation at the ankle was possibly used to reduce the forward angular momentum of the body, and coactivation at the hip to support the trunk. The younger adults in this study did not show an increased coactivation during trip recovery compared to walking. This suggests they did not require increased stiffness at the ankle and hip to reduce the forward angular momentum of the body during trip recovery and to maintain the trunk upright. In the older adults coactivation at the ankle was particularly high during elevating strategy recoveries and relatively low compared to walking in lowering strategy recoveries. This suggests that during elevating strategy recoveries they needed to increase stiffness to reduce the forward angular momentum of the body, while during lowering strategy recoveries a recovery technique was used that did not require such a high stiffness. The relatively low LRA of older adults during elevating strategies compared to younger adults, and the similar LRA of younger and older adults during lowering strategy recoveries support this suggestion (Table 6.1).

The older adults were not as successful as the younger adults in recovering with an elevating strategy and attempted to stiffen their ankle joint to reduce the forward angular momentum. The older adults were more successful in reducing the forward angular momentum of their body during lowering strategy recoveries and therefore did not require to stiffen their ankle joint as much. The older adults showed a higher coactivation at the hip during trip recovery than during walking. This was possibly as it required a higher stiffness at the hip joint to maintain the trunk in an upright position during trip recovery than during walking.

6.5. Arm movement

This section addresses the third research question "What is the contribution of arm movement to successful trip recovery in both younger and older adults?".

6.5.1. Experimental results

A typical movement of a younger adult during an elevating recovery strategy (the obstructed leg is lifted over the obstacle) is shown in Figure 6.6.

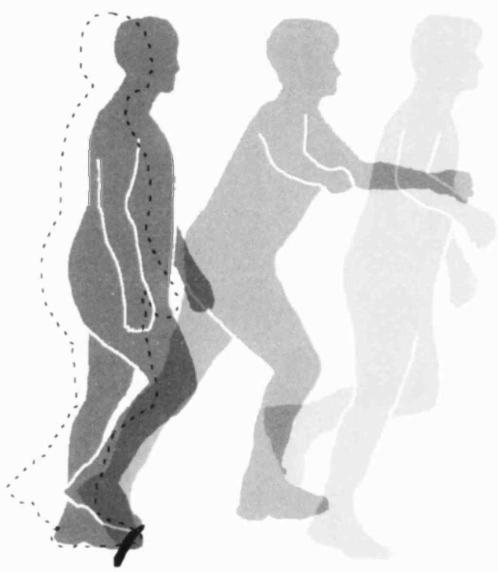


Figure 6.6 Silhouette figure of a typical elevating strategy trip recovery of a younger adult, showing arm movement during the recovery. Maximal arm displacement is shown in the middle shaded silhouette.

The younger adults moved (resultant displacement) the arm contralateral to their recovery limb more than the older adults in lowering strategy recoveries, 0.48±0.38 ALs versus 0.16±0.22 ALs ($p<0.05$) (Table 6.4). Despite apparent differences, due to large individual variations no significant differences were found in the extent of arm movement between younger and older adults during elevating strategy trip recoveries (0.22±0.20 ALs and 0.04±0.03 ALs for the younger and older groups respectively). The peak velocity of the resultant arm movement in early trip recovery was not significantly different between the younger and older adults (Table 6.4).

In comparisons between recovery types, the velocity of the ipsilateral arm was higher in lowering than in elevating strategy recoveries for the younger adults. The younger adults also showed a larger displacement of the contralateral arm in lowering than in elevating strategies (Table 6.4).

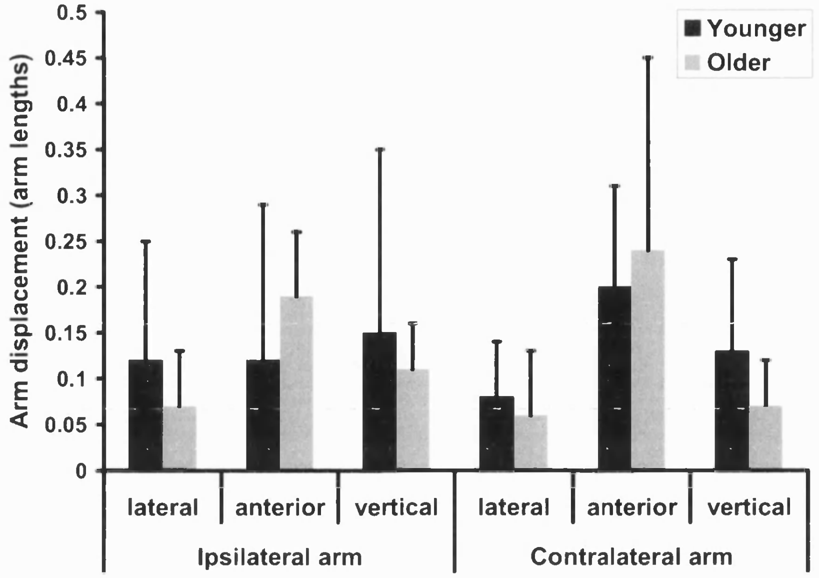
*Table 6.4 Mean maximum resultant arm CM displacement and peak resultant arm velocity during early trip recovery for younger and older adults. Significant differences (independent t-test, $p<0.05$) between the younger and older groups are indicated with *, and between elevating and lowering strategy recoveries with $^{\&}$.*

		Ipsilateral arm		Contralateral arm	
		Displacement (AL)	Velocity (AL/s)	Displacement (AL)	Velocity (AL/s)
Elevating	Younger	0.04±0.07	0.39±0.42	0.22±0.20	0.40±0.40
	Older	0.12±0.07	0.66±0.40	0.04±0.03	0.44±0.37
Lowering	Younger	0.17±0.37	0.91±0.71$^{\&}$	0.48±0.38$^{\&}$	0.58±0.49
	Older	0.14±0.18	0.72±1.24	0.16±0.22*	0.61±0.58

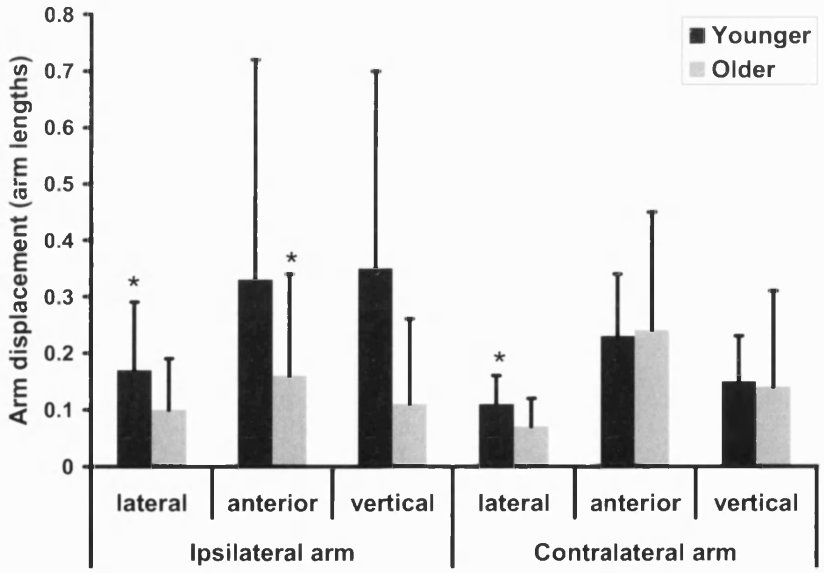
During early recovery, the vertical displacement of the whole body CM was positively correlated with vertical arm CM displacement of the arm contralateral to the recovery limb ($r = 0.665$, $p < 0.01$) for the younger adults only. Older adults did not exhibit a similar correlation between arm displacement and vertical body CM displacement ($r = 0.091$). However, body CM displacement was not significantly different between younger and older adults for both elevating (younger: 0.09±0.07 LLs and older: 0.09±0.04 LLs) and lowering recovery strategies (younger: 0.14±0.17 LLs and older: 0.15±0.09 LLs). This means there was no overall difference in body CM displacement between younger and older adults, but the trials with larger arm movement had a larger body CM displacement.

Considering the directional components of the arm CM motions, in lowering strategy recoveries the younger adults showed significantly ($p<0.05$) larger lateral displacement than older adults for both arms (0.17±0.12 and 0.11±0.05 ALs versus 0.10±0.09 and

0.07±0.05 ALs for the ipsi- and contralateral arms respectively), while anterior displacement was significantly larger in the younger adults in the arm ipsilateral to the recovery limb (Figure 6.7b). The directional components of arm movement in elevating strategy recoveries were not significantly different between the younger and older adults (Figure 6.7a). The relation between the directional components differed however for younger and older adults. The older adults showed a relatively larger anterior arm movement in relation to the vertical movement than the younger adults in early trip recovery. For elevating strategy recoveries, the vertical arm displacement of the younger adults was 125% and 65% of the anterior displacement for the arm ipsilateral and the arm contralateral to the recovery limb respectively, while they were 58% and 29% for the older adults. For lowering strategy recoveries the vertical arm displacement of the younger adults was 106% and 65% of the anterior displacement for the arm ipsilateral and the arm contralateral to the recovery limb respectively, while they were 69% and 58% for the older adults.



a



b

*Figure 6.7 Maximum arm CM displacement in lateral, anterior and vertical direction for an elevating (a) and a lowering (b) strategy recovery. Data are split in the arm ipsilateral and the arm contralateral to the recovery limb and elevating and lowering strategy recoveries. Significant differences between younger and older adults are indicated with *.*

The correlation of ERA with arm movement was investigated, because the largest influence of arm movement was expected before the recovery foot contacted the floor. In elevating strategies the magnitude of the 'Arm Recovery Amount' (ARA) of the contralateral arm was a significantly ($p<0.05$) larger percentage of ERA in the younger (13.0%) than in the older adults (-2.7%). This percentage was positive in the younger adults and negative in the older adults (Table 6.5), which means arm movement helped to reduce the forward angular momentum of the body in the younger adults while in the older adults arm movements increased the forward angular momentum. This percentage was during lowering strategy recoveries not significantly different between the younger and the older adults. ARA of the older adults was significantly ($p<0.05$) larger in elevating than in lowering strategy recoveries.

*Table 6.5 Average arm recovery amount percentage (the percentage contribution of the arms to overall body recovery amount) of the arm contralateral to the recovery limb for younger and older participants for elevating and lowering strategy recoveries. Significant differences (independent t-test, $p<0.05$) from the younger group are indicated with * and between elevating and lowering strategy recoveries with $^{\&}$.*

	Arm recovery amount percentage	
	Elevating	Lowering
Younger	13.0±12.5%	9.1±9.1%
Older	-2.7±0.6%*	7.8±6.4%$^{\&}$

6.5.2. Discussion

During both elevating and lowering strategy recoveries younger subjects exhibited the largest arm movement in the arm contralateral to the recovery limb. This arm was moved forward and lifted upwards, giving it a backward swing movement relative to the trunk. The arms were also moved laterally (outwards) during early trip recovery. The displacement was larger in the arm contralateral to the recovery limb to counter-balance the motion of the recovery limb in a similar way as can be seen by arm swing in walking or running. Laterally, the arms initially moved away from the body before returning to their original position. This lateral displacement would increase the moment of inertia about the frontal axis during early recovery and provide extra lateral stability by minimising rotations in this plane during the initial phases of the recovery. This suggests that the younger adults used a coordinated arm movement to help maintain balance and to assist in actual recovery from the trip situation.

Vertical arm movement was positively correlated with the body centre of mass displacement in elevating strategies of the younger adults only ($r = 0.665$). This means that trials with the largest extent of arm movement were associated with increased elevation of the body CM. This elevation of the body CM from greater arm motion will provide extra time for placement of the recovery limb. The body CM displacement during early trip recovery was however not significantly different between the younger and older adults. This means that while there was no overall difference in body CM displacement between younger and older adults, the trials with larger arm movement had a larger body CM displacement.

Arm movement also appeared to play a role in reducing the forward angular momentum of the body during elevating (13.0%) and lowering strategy recoveries (9.1%) in the younger adults. The contribution of the arm movements for older adults was 7.8% for lowering strategies and -2.7% for elevating recovery strategies. Therefore, a relative backward motion of the arms can be seen to make an important contribution to the reversal of the induced angular momentum during early trip recovery (prior to contact of the recovery limb). These contributions will support the contributions made by the initial support limb as demonstrated previously by Pijnappels et al. (2004; 2005a; 2005c).

The younger adults moved the arm contralateral to the recovery limb more than the older adults in both elevating and lowering strategies. This difference was mainly in the vertical direction in elevating strategy recoveries and in the lateral direction in lowering strategy recoveries. The older adults exhibited a lower ratio of vertical to anterior arm displacement, resulting in a more reaching arm movement, while the younger adults had a lifting arm movement, in both elevating and lowering strategy recoveries. This agrees with findings by Allum et al. (2002) in perturbations induced during quiet stance. The more reaching movement of the older adults suggested that they were anticipating an unsuccessful recovery and reached forward to protect themselves against a potential fall rather than using their arms to make an attempt to recover balance. It has to be emphasised that the younger and older adults had similar scores on the fear of falling questionnaire and showed no presence of fear of falling.

It can be concluded that arm movements play a contributing role in trip recovery in younger adults by elevating the body CM and reducing the forward angular momentum of the body, both providing more time for appropriate positioning of the recovery limb. The older adults showed a more protective arm response, reaching forward to arrest a possible fall, which limited the contribution which could be made by the arms to raise body CM and reverse angular momentum.

6.6. Range of motion

This section investigates the fourth research question "What is the difference in the utilised range of motion of the joints (RoM) of the lower limb between younger and older adults, and how does this utilised RoM influence trip recovery?".

6.6.1. Experimental results

It was investigated how the RoM utilised during trip recovery differed between younger and older adults, and between trip and walking trials. Also, how utilised RoM correlated with recovery amount and the size of the perturbation was determined. Joint angle trajectories of representative trials for younger and older adults are shown in Figure 6.8.

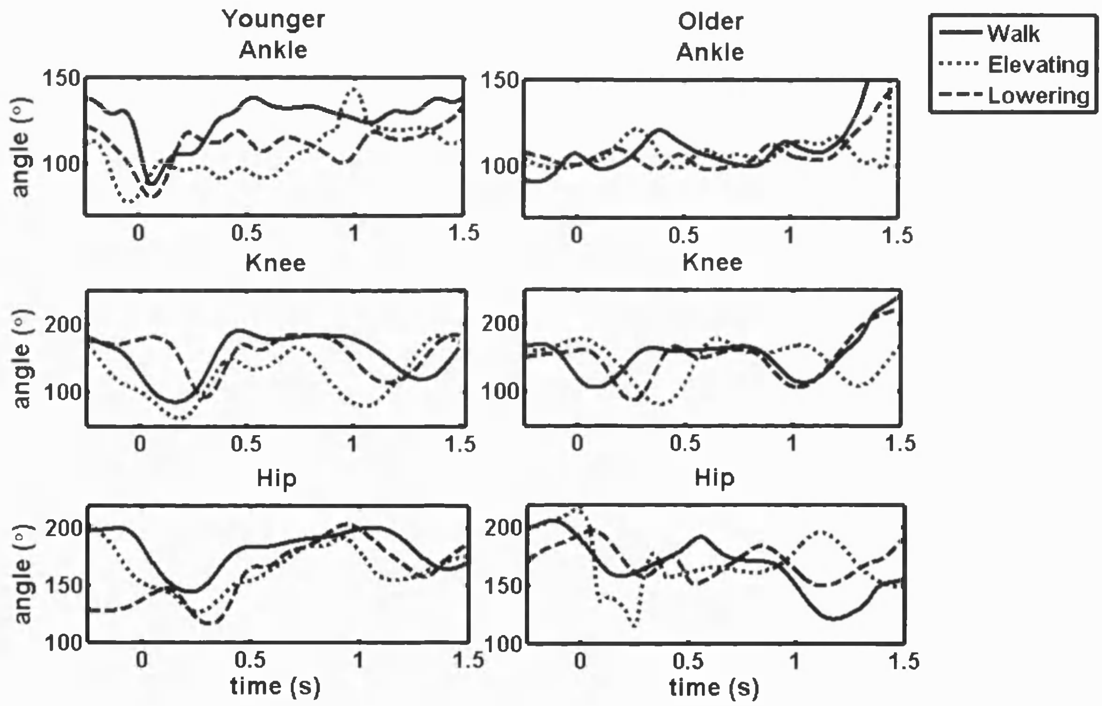


Figure 6.8 Ankle, knee and hip joint angles of typical trials from younger and older adults.

The utilised RoM of the ankle was significantly ($p<0.05$) smaller in older than in younger adults during walking and elevating strategy recoveries. The utilised RoM of the knee and hip were not significantly different between younger and older adults. Both the younger and the older adults used a significantly ($p<0.05$) larger RoM at the knee and hip of the recovery limb during lowering strategy recoveries than during walking (Table 6.6). The older adults also used a larger RoM at the ankle during lowering strategy recoveries than during walking. The utilised RoM of the younger adults at the knee and hip was significanlty larger ($p<0.05$) during lowering strategies than during elevating strategy recoveries, while for older adults the utilised RoM of the ankle and knee were significantly larger during lowering than during elevating strategy recoveries (Table 6.6).

The correlation of ERA with utilised RoM was investigated as it was expected that a larger ERA would provide more time for placement of the recovery limb and therefore coincide with a larger utilised RoM during trip recovery. To investigate the effect of utilised RoM on recovery amount the correlation of this variable with LRA was investigated.

The utilised RoM of both the ankle and hip of younger adults were positively correlated ($p<0.01$) with ERA during elevating strategy recoveries. This means they reduced a larger amount of their forward angular momentum during early trip recovery when a larger RoM was utilised at the ankle and hip. The utilised RoM of the ankle was negatively correlated ($p<0.01$) with LRA during lowering strategy recoveries for older adults. This means they were able to reduce a larger amount of their forward angular momentum when a smaller RoM was utilised.

*Table 6.6 Average utilised RoM of the ankle, knee and hip of the recovery limb for younger and older subjects, during walk trials and for elevating and lowering strategy recoveries. Significant differences to younger subjects ($p<0.05$) are indicated with *, to walk trials with $^{+}$, and to elevating strategy recovery trials with $^{\&}$.*

		Utilised RoM recovery limb (°)		
		Ankle	Knee	Hip
Younger	Walk	40±19	78±13	53±23
	Elevating	38±21	80±17	53±17
	Lowering	40±16	95±19$^{\&+}$	69±16$^{\&+}$
Older	Walk	27±11*	75±11	53±15
	Elevating	25±11*	74±9	63±20
	Lowering	36±14$^{\&+}$	87±15$^{\&+}$	67±19^{+}

6.6.2. Discussion

The passive RoM of joints reduces when adults get older (Shephard, 1997). During trip recovery a person does not utilise the full range of this passive RoM. During stepping down movements older adults tend to use a larger percentage of their passive ankle and knee RoM than younger adults (Lark et al., 2004). Wojcik et al. (2001) however found no difference in joint RoM utilised during recovery from sudden release from a forward lean between a younger and an older subject group. In the present study only the utilised RoM of the ankle during walking and elevating strategy recoveries was significantly smaller for older than for younger adults.

The utilised RoM of the ankle of the recovery limb of older adults was negatively correlated with LRA during lowering strategy recoveries. This means using a smaller RoM of the recovery limb implied a larger recovery amount (larger reduction in forward angular momentum induced by trip). This can be explained by assuming the body has pendular movement, if the forward angular momentum is reduced by a larger amount there will be less rotation at the ankle joint. A positive correlation was found between the used RoM of the recovery ankle and hip of younger adults and ERA during elevating strategy recoveries. This means when they reduced a larger amount of their forward angular momentum during early trip recovery a larger RoM was utilised at the ankle and hip. Although the utilised RoM of the younger and older adults were similar during trip recovery (except from the ankle RoM utilised during elevating strategy recoveries) their utilised RoM were correlated in a different amount with recovery amount. This indicates that older adults were not able to reduce their forward angular momentum in similar amounts as younger adults when utilising similar RoM and again might suggest the use of a different recovery technique while using the same recovery strategy.

6.7. Recovery step length

Section 6.7 investigates the fifth research question "How does the recovery step length vary in relation to trip recovery strategies in both younger and older adults?".

6.7.1. Experimental results

The overall, medio-lateral and anterior-posterior recovery step length (defined as the distance between the ankle coordinates of the obstructed foot at contact with the tripping device and at contact of the recovery leg with the force plate) are presented in Table 6.7.

In walking trials only medio-lateral recovery step length was significantly ($p<0.05$) smaller for older than for younger adults. The younger subjects showed a significantly ($p<0.05$) larger overall, medio-lateral and anterior-posterior recovery step length during elevating strategy recoveries than during a walk and a larger overall and anterior-posterior recovery step length during lowering strategy recoveries than during walking. Older adults did not show any significant difference in recovery step length during elevating or lowering strategy recoveries and walking. The younger subjects showed a significantly larger ($p<0.05$) medio-lateral recovery step length during elevating than lowering strategy recoveries. A larger lateral recovery step length can be expected to provide better lateral stability. Overall, medio-lateral and anterior-posterior recovery step length of older adults were significantly lower ($p<0.05$) than those of younger adults in elevating strategy recoveries. In lowering strategy recoveries overall and anterior-posterior recovery step length of older adults were significantly ($p<0.05$) smaller than in younger adults.

A positive correlation was found between total response step time (defined as time between contact with tripping device and first contact with tripping device) and overall recovery step length ($p<0.01$) for older adults during both elevating and lowering strategy recoveries and for younger adults for elevating strategy recoveries only. This means that overall recovery step length was larger when more time was available to place the recovery limb.

Both overall and anterior-posterior recovery step length were positively correlated with the response time in the response test (data for older adults only) during lowering strategy recoveries, while lateral recovery step length was negatively correlated ($p<0.05$) during elevating strategy recoveries. This means that older adults with a fast response in the response test had smaller overall and anterior-posterior recovery step length during lowering strategy recoveries and a larger lateral recovery step length during elevating strategy recoveries.

It would be expected that the recovery step length is influenced by ERA, as a large ERA would provide more time for placement of the recovery limb and allow for better placement of the recovery limb. The recovery step length itself would be expected to influence LRA, as a recovery step length closer to the optimal recovery step length results in a larger reduction of the angular momentum of the body.

For both the younger and the older adults overall recovery step length was positively correlated with ERA and anterior-posterior recovery step length was negatively correlated with ERA during elevating stratery recoveries. This means that a larger reduction of the angular momentum in early trip recovery went together with a larger overall, but shorter recovery step length. For the younger adults medio-lateral recovery step length was also positively correlated with ERA during elevating strategy recoveries. This means the younger adults used a wider recovery step when the angular momentum was reduced more. During lowering strategy recoveries recovery step length was not correlated with ERA. During both elevating and lowering strategy recoveries recovery step length was not correlated with LRA. This indicates that another mechanism than simply increasing recovery step length might also reduce the forward angular momentum. When recovering from a trip with a large recovery step length a large extensor moment is required at the knee to reduce the forward angular momentum. It may for some individuals be more beneficial to recover with a smaller recovery step length, which does not require such a large knee extensor moment and where the forward angular momentum can be reduced by an ankle plantarflexor moment.

*Table 6.7 Mean overall, medio-lateral and anterior-posterior recovery step length (relative to LL) for younger and older subjects for walk trials and elevating and lowering recovery strategy trials. Significant differences to younger subjects ($p<0.05$) are indicated with *, to walk trials with $^{+}$, and to elevating strategy recovery trials with $^{\&}$.*

		Overall recovery step length (LL)	Medio-lateral recovery step length (LL)	Anterior-posterior recovery step length (LL)
Younger	Walk	0.74 ± 0.12	0.21 ± 0.06	0.70 ± 0.13
	Elevating	$0.84 \pm 0.22^{+}$	$0.24 \pm 0.05^{+}$	$0.81 \pm 0.23^{+}$
	Lowering	$0.84 \pm 0.24^{+}$	$0.20 \pm 0.08^{\&}$	$0.82 \pm 0.24^{+}$
Older	Walk	0.68 ± 0.17	0.16 ± 0.05*	0.66 ± 0.18
	Elevating	0.64 ± 0.12*	$0.20 \pm 0.06^{*+}$	0.61 ± 0.11*
	Lowering	0.70 ± 0.26*	0.17 ± 0.08	0.67 ± 0.27*

6.7.2. Discussion

Typically older adults have a gait with shorter step length and a widened base of support (Payne & Isaacs, 1987; Spirduso, 1995; Shephard, 1997) or similar step width (Blanke & Hageman, 1989; Owings & Grabiner, 2004). A widened base of support provides better stability, as this makes it easier to retain the CM within the base of support. As balance is more challenged in trip recovery than walking, a larger step length was expected in trip recovery trials. Recovery step length was expected to be increased mainly in anterior-posterior direction as this would be beneficial in the reduction of the forward angular momentum of the body. A larger anterior-posterior recovery step length could be accompanied with a larger medio-lateral recovery step length to maintain lateral stability.

The younger adults showed an increased overall recovery step length during elevating and lowering strategy recoveries compared with walking (Table 6.7). Older adults only increased their medio-lateral recovery step length during elevating strategy recoveries compared with walking (Table 6.7). This suggests the younger adults increased their step length and step width to recover more successfully from a trip. The older adults may not have been able to increase step length because of too a slow response, too slow a movement velocity, or insufficient lower limb strength. As the recovery step length in older adults was not correlated with response time in the response test and it was shown in section 6.3 that younger and older adults were not significantly different in peak joint moments, it was assumed that movement velocity was the main limiting factor for recovery step length in older adults. The older adults had a significantly ($p<0.05$) smaller medio-lateral recovery step length than the younger adults during walking, which does not agree with the earlier mentioned findings in literature that older adults have a widened recovery step during gait. Also during elevating strategy recoveries the older adults had a significantly ($p<0.05$) smaller medio-lateral recovery step length than the younger adults; this could be caused by a slower response time and movement velocity of older adults, which would leave them less time for placement of the recovery limb and would therefore result in a smaller recovery step. Another explanation would be that the older adults were not capable to produce a large enough knee support moment that would be required to recover with a larger step.

The older adults did not show a significant difference in recovery step length during elevating or lowering strategy recoveries, which is in agreement with the findings of Pavol et al. (2001). A positive correlation was found between total response step time (defined as time between contact with tripping device and first contact with tripping device) and total recovery step length ($p<0.01$) and anterior-posterior recovery step length ($p<0.01$). This means a larger recovery step occurred when there was more time to place the leg forward. Both overall and anterior posterior recovery step length during lowering strategy recoveries were positively correlated with the response time in the response test (data for older adults only), while lateral recovery step length during elevating strategy recoveries was negatively correlated ($p<0.05$). This indicated adults with a slower response in the response test showed a larger anterior-posterior and smaller lateral recovery step length during trip recovery. This means the older adults with a slower response had a longer and narrower recovery step in relation to older adults with a faster response. This does not agree with what would be expected; it was expected that people with a faster response time would have more time to place their recovery limb and therefore have a larger recovery step. It was expected that recovery step length would be mainly increased in the anterior-posterior direction to be able to better reduce the forward angular momentum of the body and in a smaller amount medio-laterally to provide lateral stability. This finding is difficult to explain, possible causes could be that the older adults in this study were all healthy and they did not have a slowed response and loss of sensation to such a degree that this would prevent them from increasing their recovery step length.

For both younger and older adults the overall recovery step length was as expected positively correlated with ERA. This means they used a larger recovery step when the forward angular momentum was reduced more during early recovery, which would provide more time for placement of the recovery limb and allow for a larger recovery step. During lowering strategy recoveries recovery step length was however not correlated with ERA. The recovery step length was expected to influence LRA, as a recovery step length closer to the optimal recovery step length results in a larger reduction of the angular momentum of the body. During both elevating and lowering strategy recoveries recovery step length was however not correlated with LRA. This indicates that another mechanism than simply increasing recovery step length was also used to reduce the forward angular momentum. When recovering from a trip with a large recovery step a large extensor moment is required at the knee to reduce the forward angular momentum. It is possible that for some individuals it is more beneficial to recover with a smaller recovery step, which does not require such a large knee extensor moment and where the forward angular momentum can be reduced by an ankle plantarflexor moment.

6.8. Summary of the contributions to successful trip recovery

This section summarises the mean findings in relation to the five research questions.

Question 1: What is the contribution of the recovery limb to successful trip recovery in both younger and older adults?

Hypothesis: Younger adults have a more rapid muscle response and coordinated movement of the recovery limb to allow placement of the recovery limb to assist trip recovery. As maximum muscle force decreases with age, it was expected that younger adults would show larger joint moments in the recovery limb during trip recovery allowing a larger reduction of the forward angular momentum of their body than older adults.

The contribution of the recovery limb differed between younger and older adults for elevating strategy recoveries only. The younger adults were able to reduce a larger amount of their forward angular momentum during elevating strategy recoveries than older adults, while this amount was similar during lowering strategy recoveries. The younger adults increased their support moment during elevating strategies relative to walking. The older adults did not do this, which is likely causing them to be less successful in using elevating strategy recoveries. This lower support moment of older adults during elevating strategy recoveries appears to be mainly caused by a smaller contribution of the knee to the support moment. This smaller contribution of the knee to the total support moment can either be caused by a smaller peak torque at the knee, or due to placement of the limb at ground contact, which might not allow for sufficient moment production.

Question 2: How do muscle sequencing and coactivation influence successful trip recovery in both younger and older adults?

Hypothesis: During trip recovery older adults will exhibit higher muscle coactivation than younger adults and muscle sequencing will differ between younger and older adults.

The older adults showed a consistent muscle sequencing response, which was similar for all strategies, while the younger adults showed varying muscle responses depending on the recovery strategy used; indicating younger adults had more adaptable movement strategies available. Both the younger and older adults showed relatively large coactivation at the ankle and hip during trip recovery in order to stabilise trunk movement during trip recovery. Coactivation at the ankle would reduce the forward angular momentum of the body and this was evident in both younger and older adults during lowering strategy recoveries, but only in older adults in elevating strategy recoveries. The results demonstrate some increased tendency for older adults to use coactivation during trip recovery, but the hypothesis that older adults would exhibit overall higher coactivation during trip recovery than younger adults was not fully supported.

Question 3: What is the contribution of arm movement to successful trip recovery in both younger and older adults?

Hypothesis: Younger adults use their arms more effectively than older adults, due to an increased range of motion and generation of opposite angular momentum.

In younger adults arm movements play a fall preventive role during trip recovery, by contributing to the elevation of the body centre of mass and by decreasing the forward angular momentum of the body, which both provide more time for placement of the recovery limb. Older adults used their arms in lowering strategy recoveries only. In elevating strategy recoveries the older adult showed a more protective movement, anticipating a possible fall.

Question 4: What is the difference in joint range of motion of the lower limb between younger and older adults, and how does this range of motion influence trip recovery?

Hypothesis: Older adults use a smaller range of motion of their lower limbs than younger adults and that this limits their trip recovery success.

Utilised RoM differed between younger and older adults during trip recovery. The range of motion of older adults was restricted at the ankle joint, while a larger RoM was utilised at the hip joint. This suggests younger and older adults used different recovery techniques within the same recovery strategy (elevating vs. lowering).

Question 5: How does the recovery step length vary in relation to trip recovery strategies in both younger and older adults?

Hypothesis: Older adults are not able to utilise a recovery step length as large as younger adults, and this limits trip recovery success.

The younger subjects showed a larger overall, medio-lateral and anterior-posterior recovery step length during elevating strategy recoveries than during a walk and larger overall and anterior-posterior recovery step length during lowering strategy recoveries than during walking. Older adults did not show any significant differences in recovery step length during elevating or lowering strategy recoveries and walking. This suggests that the younger adults were able to put their recovery limb in a better position to reverse the forward angular momentum of the body than the older adults were.

Chapter 7: General discussion

The research questions posed in Chapter 1 could all be answered with the experimental results (Chapter 6). The issues that arose from these results will be discussed in this chapter. Both the experimental and the modelling research approach of this thesis had some limitations, which will be discussed in this chapter followed by directions for future research originating from this thesis.

7.1. Discussion of main findings

It was found that older adults adopted a lowering strategy more often than younger adults, which agrees with the findings of Pijnappels et al. (2005a). To understand what causes this, the difference between elevating and lowering strategy recoveries has to be understood and are therefore described again briefly. Elevating strategies are used in response to perturbations in the early and mid swing phase and lowering strategies in response to perturbations in the late swing phase. In an elevating strategy the obstructed leg is lifted over the obstacle, while in a lowering strategy the body CM is too close to or in front of the CoP which makes it difficult to lift the obstructed limb over the obstacle and therefore this limb is put to the ground prior to the obstacle and the contralateral limb becomes the recovery limb.

There are different possible explanations for why older adults are less effective in elevating strategy recoveries than younger adults. One possible framework to explain these differences is to consider that successful recovery using an elevating strategy requires the recovery limb to be used in an energy absorbing manner in order to reverse the destabilising angular momentum. In individuals (e.g. older adults) not capable of using this strategy it may be that the recovery limb is used more as a strut than a spring. Many findings of this study would seem to reinforce this framework. This theory assumes that younger adults absorb energy at the knee during trip recovery, which was suggested by their large extension moments at the knee of the recovery limb, which in combination with the large angular velocity suggests a high negative knee power. Similar energy absorbing and pivoting strategies have been found in perturbation of gait of animals, depending on the phase of the gait cycle at which the perturbation occured (Daley & Biewener, 2006). The older adults could possibly not produce a high enough joint moment to use this absorbing strategy, or their movement velocity and response time were not sufficient to place their recovery limb in the right position to absorb energy. It is suggested that they used a pivoting strategy, in which the recovery limb is used as a strut and the movement of the body can be described by pendular motion with a rotational spring (ankle stiffness) at the base of the pendulum. The observed high coactivation (Table 6.3) may be responsible for the higher limb stiffness required of the limb to act as a strut. It is suggested that only the younger adults were capable of using an energy absorbing trip recovery strategy and that it was only required to use this strategy during elevating strategy recoveries. During lowering strategy recoveries the obstructed foot is set down immediately after trip onset and already absorbs some energy.

The more frequent use of a lowering strategy recovery by older adults might be due to their slower response time and lower muscle strength. To recover with an elevating strategy in mid to late swing can be expected to require a fast and powerful muscle response, as the swing limb has to be lifted over the obstacle while the body is moving towards the obstacle. The body centre of mass will be in front of or close to the centre of pressure and when the foot is obstructed it will be easier, and might feel like a 'safer option' to put this leg down to support the body; lifting the leg over the obstacle will require a quick response and higher forces in the stance limb. In particular the energy absorbing strategy, which the younger adults seem to show during elevating strategy recoveries, would require high forces at the knee and a fast response time and movement velocity to place the recovery limb in the optimal position at ground contact. The muscle onset timings found in this thesis were similar between the younger and the older adults for both elevating and lowering strategy recoveries (Figure 6.5). This suggests that trip recovery success in older adults was limited more by movement speed than by reaction time.

Both the younger and the older adults showed a higher coactivation at the ankle during elevating than during lowering strategy recoveries (Table 6.3). This indicates that they attempted to stiffen their ankle joint and attempted to reduce the forward angular momentum of their body. A higher stiffness of the ankle joint would be expected in a pivoting recovery strategy rather than in an energy absorbing strategy. The older adults showed similar ankle, knee and hip peak moments to the younger adults (Table 6.2), but muscle coactivation at the ankle was on the other hand in elevating strategy recoveries higher in the older adults. Older adults might not have been able to produce sufficiently high joint moments to adopt an energy absorbing strategy. This increased muscle coactivation can be expected to increase energy expenditure, this energy is however not wasteful as it might enhance joint stiffness (Hasan, 1985). During trip recovery it is contended that it would be more important to adopt a fall preventive instead of an energy efficient strategy, something which contradicts Forner-Cordero et al. (2005).

LRA was found to be significantly ($p<0.05$) lower for older than for younger adults during elevating strategies (0.004 vs. 0.011 m/s), while it was similar during lowering strategies (0.009 vs. 0.006 m/s) (Table 6.1). This suggests older adults were less effective in recovering with elevating strategies than younger adults. The younger adults absorbed more energy during trip recovery at the knee of the recovery limb and reduced their forward angular momentum effectively, while the older adults could not reduce angular momentum that much by their pivoting strategy in which the stiffness of the ankle reduced most of the forward angular momentum. Forner-Cordero et al. (2005) investigated the segmental energy during trip recovery in younger adults and found this was higher in lowering than in elevating strategies. This agrees with the younger adults using an absorbing strategy, as the knee absorbs energy and therefore does negative work which would result in a lower segmental energy.

The younger adults showed a larger recovery step during elevating than during lowering strategy recoveries, while during elevating strategy recoveries the older adults exhibited a smaller overall, medio-lateral and anterior-posterior recovery step length than younger adults (Table 6.7). This smaller recovery step of older adults during elevating strategy recoveries might be a possible cause of why older adults were less effective in elevating than in lowering strategy recoveries. A smaller recovery step would also be expected in a pivoting strategy than in an absorbing strategy.

During elevating strategies younger adults showed a larger LRA (reduction of angular momentum by the recovery limb) than ERA (reduction of angular momentum by the initial stance limb), while they were similar during lowering strategy recoveries (Table 6.1). This indicates that during lowering strategy recoveries an energy absorbing strategy was not needed. The older adults however showed similar ERA and LRA during elevating strategy recoveries, while during lowering strategy recoveries LRA was larger than ERA. In general, the role of the recovery limb was at least as important as that of the initial stance limb. During lowering strategy recoveries the older adults were able to recover total angular momentum to the same extent as the younger adults were.

Another explanation for the use of a pivoting strategy by the older adults, in which there is high muscle coactivation, is that they had a more panicked or unanticipated response. The high muscle coactivation at the ankle, which possibly led to the pivoting strategy, could have been a result of panic. This is in agreement with findings by Besier et al. (2003), who found that athletes showed higher activation and coactivation during unplanned than during planned movements, and a more selective activation during planned movements. The panicked response by older adults was supported by the finding that they showed a more protective arm movement, to prevent an eventual fall, while the younger adults used their arms effectively in trip recovery to raise body CM and reverse angular momentum.

The older adults had a significantly ($p<0.05$) slower walking velocity than the younger adults (1.11 vs. 1.22 LL/s), which made it more difficult to maintain balance following a perturbation, due to step frequency (Alexander, 1992). It was expected that an elevating strategy has higher balance demands, as balance has to be maintained while the initial stance limb stays on the ground and the recovery limb is lifted over the obstacle, while in a lowering strategy the perturbed limb is put on the ground providing extra time and support to regain balance early in the recovery. Since from a balance perspective it can be assumed to be easier to lower the obstructed leg to the ground instead of lifting it over the obstacle following a perturbation (especially in the later swing phase as the CM is in front of the CoP), it might be that for older adults the transition from using lowering instead of elevating strategy recoveries occurs earlier in swing than for younger adults. This was supported by the findings of Pijnappels et al. (2005a), where in some trials older subjects used lowering strategies in response to mid-swing obstructions, when elevating strategies were expected. This was not investigated directly in this study, but it was assumed that, in the trip-experiment in this thesis, trips were randomly distributed between early, mid and late swing obstructions as trips were induced randomly.

Inter-individual variation in biomechanical outcome measures was high in this study as can be seen by the high standard deviations found in the experimental data. This agrees with findings by Pijnappels et al. (2004) who found substantial between subject variation in the reduction of the forward angular momentum of the body by the initial stance limb. Not all their subjects were able to reduce the angular momentum during the push off phase of the stance limb. This high variation might make it difficult to find a general optimal trip recovery strategy and the optimal strategy might differ for each person and for each situation as trip stimuli differ in real life. The computer simulation model of trip recovery will be a helpful tool to identify individualised optimal trip recovery strategies and to identify how trip recovery success could be improved for individuals.

7.2. Limitations

It was attempted in the experimental setup to induce trips similar to trips in real life and to keep measurement errors as small as possible. The simulation model was kept as close as possible to reality, without making it overcomplicated. However both the trip recovery experiment and the trip recovery simulation model possess some limitations, which are described in this section.

7.2.1. Tripping protocol

The participants were informed that trips would be induced prior to the experiments and the tripping device was shown. This meant that the approximate position of the obstacle was known, which limited the surprise element. Trips were however induced in random trials and randomly to the left or right foot. The modified glasses and music on the earphones prevented the subjects from noticing when the tripping device was activated. Pijnappels et al. (2001) showed that after forewarning of tripping the walking pattern changed only minimally. This suggests tripping responses would also be minimally influenced by forewarning of tripping. The glasses with the lower half of vision obscured influenced the vision of the participants, while vision has been shown to be important in maintaining balance, especially in older adults (Chen et al., 2005; Poulain & Giraudet, 2005). This influence of altered vision was however similar in all participants as the same pair of glasses was used. The surprise element of not being able to see when the tripping device was activated was chosen above the fact that the glasses influence normal vision. The subjects wore an ankle support as advised by the ethics committee; this may have influenced the ankle motion, although the support was a wrap around support that influenced ankle movement minimally, as most studies investigating ankle supports found their effect on performance to be minimal (Bot & van Mechelen, 1999).

7.2.2. Participants

The participants in this thesis were all healthy and had no history of falls. As a result of this some of the main risk factors for falls may have been absent. It is however ethically difficult to put people with a history of falls through the trip recovery experimental protocol, due to fear of falling. The participants were safely secured in a harness that prevented them from falling. This may have given the participants extra confidence and influenced their trip recovery kinematics. It has however been shown by Pavol et al.

(1999b) that wearing a safety harness during tripping protocols has almost no effect on normal gait.

The number of participants in the trip recovery experiment was relatively small and a larger number of participants would have likely increased the chances of statistical differences being found. This number was however based on statistical calculations using the best data available prior to the experiments. Given the involvement of elderly and possibly frail participants the preference was not to test too many participants. The percentage of trials analysed was also relatively small. This was mainly due to the nature of the experiment; not every attempted perturbation resulted in a trip and not every trip resulted in a clear placement of the recovery foot on the force plate. The experiments were already intensive and lengthy and increasing the number of trials might have produced fatigue effects, and also increased the risk of habituation to the trip response.

7.2.3. CODA and high speed video data

The CODA markers were attached directly to the skin, the triads or the extensions of the Codamotion segmental gait analysis. Movement of the skin and the extensions relative to the skin can cause error in the calculated joint centre positions. Movement of the markers can be expected especially after impact and the error caused by this was further reduced by smoothing the data. The markers were placed, when possible, on bony landmarks to minimise skin movement. As this error would be of a relatively high frequency and body movement has a relatively low frequency, it was assumed most of the skin movement artefact was removed by the smoothing. Gittoes (2004) showed in an experimental study of impact landing that the effects of using active markers instead of lower mass skin markers are minimal (0.001 m) on the location of body landmarks.

Another error in the joint centre positions may be, as with all gait experiments, in the estimation of the joint centre positions themselves. The positions of the joint centres were derived from the marker positions and as the markers were positioned by the same person in all trials, the errors in the locations can be expected to be similar in all trials.

CODA and high speed video could not be collected simultaneously as the high speed video required extra light which interfered with the CODA system. It was therefore chosen to do the high speed video recording in separate trials. This increased the number of trials required to get a full data set. This set-up was beneficial in some respects since the simulation model evaluation trials were independent from those in which the spring-damper parameters were derived.

7.2.4. Spring and damper parameters of the foot

The spring and damper parameters of the foot were defined using digitised ankle, metatarsal and toe coordinates from high speed video data and ground reaction force data. The horizontal and vertical spring and damper parameters were estimated by matching the force and displacement data using the Downhill Simplex method. The spring and damper parameters were used in the trip recovery model where the foot segments were assumed to be rigid, while the coordinates were from a real foot which is not rigid during ground contact. There was a considerable within-subject variation in the optimised spring and damper parameters. This variation was expected to be caused by variations in impact velocity and loading rate, which would modify the spring and damper properties of the foot (Aerts et al., 1995). For this reason it was chosen to use a combination of stiffness and damping coefficient values from a single trial instead of mean values.

7.2.5. Activation parameters

The parameters for the nine parameter function defining the torque-angle-angular velocity relationship were derived from literature and not subject specific. More accurate subject specific parameters could have been obtained using isokinetic dynamometer measurements. It was however decided that maximum muscle strength measurements on a dynamometer would be too strenuous for the older participants, in addition to the trip recovery protocol, and values from literature would give a good estimate as the simulation model is a simplification of reality anyway. Dynamometer measurements were done with one young subject and the nine parameters were derived and compared with those obtained from literature. The differences were deemed acceptable, which justified the use of the values derived from literature for the torque-angle-angular velocity profiles.

7.2.6. Model development

In the development of the computer simulation model simplifications were made. The aim was to keep the model as simple as possible while being able to obtain reliable and accurate outcomes. The model represented the body by rigid segments and the upper body by a single mass that could move relative to the pelvis. Actions of several muscle groups were combined as joint torques. The feet were represented by rigid segments with spring and damper systems at three different positions only. The feet in the model consisted of two segments at a fixed angle to each other, whereas in a real foot this angle is not fixed. However, the rolling movement of the foot was still present due to the fixed angle of the two foot segments. The model was personalised for one subject by using the subject specific inertia parameters and anthropometric values as input.

7.2.7. Model evaluation

While the obtained results were deemed acceptable, it proved difficult to match both kinetic and kinematic data in the model evaluation, mainly due to the fact that the model was a simplification of reality. In particular, the ground reaction forces were difficult to match. Difficulties occurred at the initial part of the model where the stance foot was already in contact with the ground. No force data were acquired for this part of the movement and initial spring displacement could not be derived accurately from the experimental data. The use of a second force plate would have given the force of this initial stance phase and could have made the evaluation more accurate. There was however only one force plate available and therefore the force of the recovery step was measured to be able to investigate the joint moments during this step. Better results could possibly be obtained with more realistic variable boundaries and by reducing the number of variables in the optimisation.

7.3. Directions for future research

This thesis was able to answer all the research questions posed in Chapter 1. However some questions were raised that require further research.

It was found, in agreement with other studies, that older adults preferred a lowering above an elevating strategy. Some possible underlying reasons for this were suggested based on the outcomes of the trip recovery experiment and by assuming that younger adults used an energy absorbing strategy while older adults used a pivoting strategy. Further research is however required to show the exact underlying reasons and to provide more evidence for the energy absorbing versus pivoting strategy theory. Further research is also required to investigate whether the older adults benefit from their preference of a lowering strategy recovery, or if it would be better for them to use an elevating strategy recovery in situations where younger adults do so. This can be investigated in the future, using the trip recovery simulation model.

Angular and vertical movements of the arms were shown to assist trip recovery in younger adults. The role of lateral arm movements on lateral stability during trip recovery needs further investigation. Further research is also required to determine whether it will be beneficial for older adults to use the more fall preventive lifting arm movements like younger adults do, or if they will be better off using the more protective reaching strategy, to prevent fall injury.

As the older adults produced similar joint moments during trip recovery to younger adults and younger adults recovered better, it can be questioned whether muscle strengthening exercises are the right approach in fall-prevention. Would it be better to focus fall-prevention on other things such as response time, confidence, flexibility and recovery technique? In older adults the ankle, knee and hip joints contributed in different amounts to the support moment during trip recovery compared with younger adults. This suggests younger adults used a different technique than older adults, while using the same overall recovery strategy. Further research is required to investigate if this different technique would be beneficial to older adults.

Similarly to previous research, a high inter-individual variation was found in the experimental data. Further research is required to investigate the reasons for this high inter-individual variation and to investigate why some adults of similar age, anthropometric and physical characteristics were able to recover from trips more successfully than others.

The first phase (prior to trip stimulus) of the trip recovery model has been evaluated. The second phase of the model evaluation is ongoing. To improve the response of the swing ankle to the trip stimulus, passive torques will be applied to this joint also at neutral joint angles. After model evaluation a sensitivity analysis will be performed to investigate what effect varying the model parameters has on the simulation outcomes. In the sensitivity analysis the spring-damper parameters and the parameters for the nine parameter torque function will be varied by 5%. The effect of these parameters on the simulation outcomes will be determined by the effect they have on the joint angle, GRF and CoM RMS components of the penalty function. In the future, the model will be applied to investigate the contributions to successful trip recovery further. Simulations will be performed to investigate the influence of peak joint moments and activation sequencing on trip recovery.

The trip recovery model has the potential to incorporate the effect of arm movements on the displacement of the upper body centre of mass, by moving the upper body mass relative to the pelvis. The experimental results showed that arm movements do play a role in trip recovery of younger adults. Further research is required to determine if this movement of the upper body can successfully simulate the reduction of the forward angular momentum of the body due to arm movement, as seen in the experiments.

The trip recovery model can potentially be a useful tool in fall-prevention practice. After taking anthropometric measurements and estimating response time, personalised simulations can be done to predict which aspects of fall-prevention therapy will be most beneficial to an individual. In future, these simulations could be used to develop tailored exercise-based fall-prevention programmes, the effectiveness of which would be assessed in randomised controlled trials.

7.4. Conclusions

This thesis set out to identify the biomechanical contributions to successful trip recovery in both younger and older adults. It has been shown that several factors contribute to successful trip recovery; the joint torques in the recovery limb, muscle sequencing and coactivation, arm movements, range of motion of the joints of the recovery limb and length and width of the recovery step. Differences in these factors between younger and older adults caused older adults to be generally less successful in trip recovery and to be forced to use a lowering strategy recovery in situations where younger adults would use an elevating strategy recovery.

The recovery limb played an important role in trip recovery during both elevating and lowering strategies of younger adults, reducing the forward angular momentum in an amount similar to (lowering strategy) or even larger than (elevating strategy) that reduced by the initial stance limb. In lowering strategy recoveries the recovery limb played a similar role in older adults to that in the younger adults, reducing a similar amount of the angular momentum as the initial stance limb. In elevating strategies however older adults showed a lower support moment and were not able to reduce the forward angular momentum with their recovery limb to the same extent as the younger adults. This seemed to be mainly due to foot positioning rather than joint moment production. It was proposed that during elevating strategy recoveries the older adults used a pivoting strategy, whilst the younger adults used an energy absorbing strategy.

In younger adults, arm movements played a fall preventive role during trip recovery, by contributing to the elevation of the body centre of mass and by decreasing the forward angular momentum of the body, which both provided more time for placement of the recovery limb. Older adults used their arms effectively in trip recovery in lowering strategy recoveries only. In elevating strategy recoveries the older adults showed a more protective movement, anticipating a possible fall. Older adults also showed a more general muscle activation response to all trips rather than a specific one, which supports their use of a pivoting recovery strategy using a relatively high coactivation at the ankle.

The utilised range of motion during trip recovery differed between younger and older adults. The range of motion of older adults was restricted at the ankle joint, while a larger range of motion was utilised at the hip joint. This suggests younger and older adults used different recovery techniques while using the same overall recovery strategy.

When investigating the size of the recovery step, it was found that trip recovery of older adults was already restricted in the early parts of recovery, where they were unable to widen their recovery step, which further restricted them in later recovery.

All investigated variables differed in some aspects between younger and older adults, indicating that despite younger and older adults using the same recovery strategies, their recovery techniques differed. The outcomes of the trip recovery experiment suggest that the underlying causes for the different techniques used by older adults was possibly a slowed movement velocity which results in the inability to position the recovery limb to allow production of sufficient support moment to recover from a trip. Simulations with the trip recovery model would be able to point out the causes of the differences in trip recovery between the younger and older adults in more detail and would be able to investigate which aspects of trip recovery need to be improved in older adults to increase trip recovery success.

To date the first phase of the evaluation of the trip recovery model has completed successfully and the second part of the evaluation is ongoing. Prior to any future simulations a sensitivity analysis will be performed. Simulations will be done to further investigate trip recovery responses of older females and to make predictions of what improvements would be most effective in increasing trip recovery success. Future work will also be done to show the potential benefits of using the trip recovery model in the development of fall-prevention programmes.

This thesis highlighted some of the differences in trip recovery between younger and older adults and suggested explanations for these differences. The differences included joint moments and recovery step length of the recovery limb and arm movements of elevating and lowering strategy recoveries in younger and older adults. Further work will determine whether older adults would benefit from a recovery strategy more similar to that of younger adults, or if they benefit more from alternative recovery strategies.

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Appendix A Trip recovery experiment appendices

Appendix A1 Independent reviews of the trip recovery experiment

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Risk Assessment

Trip Recovery Experimental Set-Up

Completed 3 September 2004

Hazard	Person	Existing Control
1. Anyone falling while getting on or off the raised platform (approx. 20 cm) on which the experimental rig is situated	Subject or Tester	Raised platform has been extended with additional sections. Edges of platform are two-toned and have banner tape as additional visual aid to indicate edges
2. Slipping on force plate	Subject	Subject will wear rubber soled shoes as pat of experiment which increases traction/friction to minimise risk of slip. Force plate will be kept clean of any debris.
3. Wearing of safety harness	Subject	Full torso harness specifically designed for gait assistive devices (Mobility

		Research [www.litegait.com]) and for use with patients up to 120 kg. Foam padding added to harness to minimise discomfort. The harness has elastic properties (webbing) to attenuate accelerations of body segments in event of loss of balance.
4. Contact of foot with trip device	Subject	Hinged metal plate has some flexibility. Plate also padded with high-density foam to attenuate impact forces experienced by the subject.
5. Use of experimental support rig	Subject	The support rig has been custom designed and constructed by professional scaffolding company (Ashton Scaffolding, Bristol) with full knowledge of intended activities. The monorail (I-beam) system and trolley is used in construction industry and engineered for load capacities equivalent to 1000 kg, approximately a factor of 10 larger than loads expected in these gait studies. All karabiners designed for use in rock climbing and for full weight-bearing.
6. Falling on surrounding area post-trip stimulus	Subject	Safety harness can be adjusted for each subject to ensure body parts do not contact ground. Mechanical stops in rail system to ensure subject does not travel too far along rail. Foam padding secured to any scaffold material adjacent to walking path.

7. Unexpected trip stimulus	Subject	Subject fully briefed regarding experimental procedures. Full familiarisation with safety harness to ensure comfort and confidence of subjects.
8. Measurement instrumentation	Subject	All instrumentation is standard for biomechanical analysis and non-invasive. All battery packs, LED markers and surface EMG sensors attached with medical-grade adhesive tape. All wires securely attached to prevent snagging on other equipment.

Inspection Regime:

Equipment	Action	Frequency
Support Rig	All fastenings of scaffolding and I-beam checked with torque wrench	Prior to and following each data collection session
Safety harness & karabiners	Material checked for proper functioning and excessive wear and tear	Between each subject

Simon Roberts

Technical Officer / Health & Safety Liaison

Sport & Exercise Science Research Group

School for Health

University of Bath

Movement strategies in recovery from a trip

Independent External Review

Professor Julie R Steele

Biomechanics Research Laboratory

University of Wollongong, Australia

I have read with great interest the research proposal by Dr Grant Trewartha pertaining to *Movement strategies in recovery from a trip.* In the Biomechanics Research Laboratory here at the University of Wollongong, we have just completed a study which focuses on mechanisms of slips in elderly rheumatoid arthritic women. As falls in older people predominantly result from slips, trips and other losses of balance when they are engaged in their usual daily activities such as walking [1,4], focussing on strategies to recovery from a trip is extremely important, particularly knowing that falls in older people are the leading cause of unintentional injury, disability, hospitalisation and death in the world [3,4].

Suitability of the study aims

The aim of the proposed study is twofold:

(i) to investigate the biomechanical differences between recovery from a trip with an initial forward step or recovery with an initial forward step including lateral movement, and

(ii) to identify the differences in lower limb joint kinetic responses observed during different trip recovery modes.

Once a trip is initiated the most effective manner to prevent injury is to be able to recover from this trip before it results in a fall. Understanding how people recover from a trip and the effects of these different recovery strategies on lower limb loading is imperative if the investigators are able to make recommendations pertaining to movement strategies which improve recovery potential and, in turn, may lower the risk of injury following a trip.

Ability of the proposed methodology to meet the objectives

To achieve the above aims the researchers propose to biomechanically investigate the recovery response of eight young healthy females after they have been tripped. The proposed subject inclusion and exclusion criteria appear appropriate. Ideally, the researchers would examine the recover strategies employed by older women as it is this group of the population who is at greatest risk of incurring injury as a result of a trip. However, it is vital that the researchers first examine the implications of trip recovery strategies using healthy subjects to ensure that lower limb loading does not exceed values likely to cause injury in older subjects. The researchers need to statistical justify the selection of eight subjects (power analysis), although this subject number has been proven sufficient to identify differences in the gait of older women when the women were required to wear different footwear or different surfaces in our previous slip research.

The methods described by the researchers to quantify the biomechanical parameters are standard techniques used internationally in gait studies, and include an optoelectronic motion analysis system to quantify motion, force platforms to assess the ground reaction forces, and electromyographic techniques to quantify the muscle activation patterns contributing to the observed motion. The demands placed on the subjects using these proposed biomechanical techniques are no more that those placed on subjects involved in similar comprehensive gait studies around the world. We have used very similar biomechanics techniques to capture the gait of a diverse range of subjects, from young children through to the elderly.

Steps proposed to ensure subject safety

As the researchers need to elicit a trip, there is an inherent risk that a subject could suffer a fall. Although the subjects are young and healthy, there is still a risk of the subjects incurring a fall-related injury. It is therefore imperative that subjects wear a safety harness throughout the trials, as has been proposed by the researchers. In studies in which there is a heightened risk of falling, we also require our subjects to wear a harness attached to a custom-designed monorail system throughout the walking trials for safety (see Figure 1). The harness system described in the present proposal sounds very similar to the harness system that we currently use. Extensive use of this harness system in our previous studies has shown that, given sufficient familiarisation, the harness system does not impede the gait of elderly women [2]. However, it is highly effective in preventing the subjects from contacting the ground in the event of a slip or trip. As part of our familiarisation protocol we require all subjects to "fall" in the harness system and to "swing back and forth" so they are truly confident that the harness will arrest their motion in the case of a fall (as a relative "fun" activity, this also serves to relax the subjects before the gait trials start, in turn, allowing their walking action to be more natural).

Figure 1: A subject wearing the harness

In an "ideal" study investigating trip responses, the subject would be unencumbered by a harness and not informed of the impending hazard such that the trip would be totally unexpected, as usually occurs in the "real world". However, such a testing scenario would be unethical due to the high risk of injury from a fall, should the subject not recovery adequately.

Validity of the proposed trip stimulus

Trips in the "real world" usually result when there is insufficient clearance between the hazard and the walker's toe. The proposed trip stimulus (a hinged metal plate 50 mm high) will provide a valid obstacle to induce a trip and the tolerance between the ground and the toe during walking is usually less than 50 mm. The location of the trip stimulus is dictated by the need to record ground reaction forces during the trip and therefore it must be placed on the platform itself. The researchers have not specified which limb will be targeted for the trip stimulus (dominant or non-dominant). As limb dominance can affect movement response, it is recommended that the researchers either specify which limb will be tested (and therefore determine limb dominance at the beginning of their experimental protocol) or test each limb equally and account for limb dominance in the experimental design).

In summary, I support the study design and research methods proposed by Dr Trewartha. The proposed methods are ethical and represent a biomechanical approach that is accepted internationally. Should Dr Trewartha be given approval to conduct the study, I look forward to his findings as they will contribute fundamental knowledge in terms of identifying movement strategies to assist people recovering from a trip.

Yours sincerely

Julie R Steele, PhD

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6 September 2004

Movement strategies in recovery from a trip

Response to External Review

We thank Professor Steele for her review of the intended study and welcome her broad support of our experimental design and proposed methodology.

In her review, Professor Steele makes a number of important points, the principal ones we would like to address briefly here.

1. Sample size calculation

Professor Steele correctly identifies that the selected sample size requires justification. A calculation has been conducted with support from a recognised statistician and is outlined in section A51 of the main body of the ethics application form. We apologise for its omission from the experimental protocol forwarded to Professor Steele.

2. Dominant versus non-dominant limbs

Professor Steele makes another important point with respect to response strategies perhaps altering based on whether the dominant or non-dominant limb is used for the principal recovery step. It was our intention to elicit recoveries using both limbs in each subject by inducing trips on the left and right feet in a pseudo-random order. However, we had not stated this explicitly and had not incorporated explicitly dominant/non-dominant limb as an independent variable in our analysis. This has now been rectified and we thank Professor Steele for initiating this improvement in our methodology.

Paulien Roos / Grant Trewartha

Sport & Exercise Science Research Group

School for Health

University of Bath

Appendix A2 Participant information sheet younger adults

Movement strategies of younger people to recover from a trip

PARTICIPANT INFORMATION SHEET

Date: 12 July 2005 Version: 4

Dear participant,

You are being invited to take part in this research study. Before you decide it is important for you to understand why the research is being done and what it will involve. Please take time to read the following information carefully and discuss it with others if you wish. Ask us if there is anything that is not clear or if you would like more information. Take time to decide whether or not you wish to take part.

The purpose of this study is to get better insight into strategies used to recover from a trip. People with risk of falling increase their base of support when walking. This basically means they make wider steps. This is a more stable manner of walking, but will also slow people down and may cost more energy.

In this experiment we want to look at the recovery step length when people recover from a trip. We want to look at the direction and the length of the recovery step. What way of recovery is energetically optimal, but also offers good stability? An initial group of younger participants will be compared later with a group of older participants. The information we get from this study may be used as advice for therapies preventing senior people from tripping.

The total duration of this study will be about 6 months, with experiments taking place between June - September 2005. Each participant will only need to be involved for about a day (6 hours).

In this study a group of 8 to 10 females between 20 and 35 years old will be studied. Female subjects only were chosen to get a group of people with similar characteristics, and because female seniors have been reported to trip more often. Participants have to be of a maximum weight of 120 kilos, because this is the maximum weight the safety harness can support.

It is up to you to decide whether or not to take part. You will also be asked to sign a consent form. You will still be free to withdraw at any time and without giving a reason.

What will happen to you if you take part?

Initially we will send your GP a letter to inform them that you plan to take part in this experiment. The experiment itself will take about half a day (6 hours) for each participant. It will take place in the Sport and Exercise Science laboratory of the School for Health at the University of Bath. You will have to sign a written consent form before the experiment will start. After that you will get time to familiarise with the experimental set-up, and you can decide if you are still happy to continue with the experiment. After this you will be asked to walk several times over a walkway (approximately 10 meters). On some occasions, the unexpected obstruction of one foot at some part of the walkway will induce a trip. The other foot can move freely to recover from this trip. You will be asked to recover as well as possible from this trip with a single step. To make sure you cannot fall, you will be secured in a full-body safety harness attached to an overhead rail. You will be wearing glasses that obstruct you from seeing the ground to make sure the trip comes as a surprise.

To be able to analyse your motion anthropometric measurements will be taken. These are measurements of body dimensions. You will lay down on a table and body lengths, widths, depths and perimeters will be measured. We will also measure your body weight using standard scales.

To record the movement data the experiments will be video taped. There will also be some markers attached to the body with tape, which will be used to analyse our movements. The application of the markers will require you to be wearing short pants and a t-shirt. You will have to wear ankle support to reduce the risk of injury. This support will be provided to you.

All information which is collected about you during the course of the research will be kept strictly confidential. All videotapes will be kept in a locked room, which can only be accessed by the investigators. All computer data will be made anonymous. Any information about you which leaves the department will have your name and address removed so that you cannot be recognised from it.

The results of this study will be used in a three-year research PhD project on movement strategies of senior people to recover from a trip. The results will probably be published in a scientific journal. You will not be identified in any publication about this study. After the data have been analysed you will be sent a summary of the results and some personal recommendations based on our findings.

The Bath Local Research Ethics Committee has approved this study. This proposal is covered by the general compensation arrangements of the University of Bath (contact: Lisa Pritchard, University Insurance Officer, tel: 01225 6378).

If you would like to have more information about this experiment, please contact Paulien Roos (01225 384323), or Grant Trewartha (01225 383055).

Thank you for reading this,

Paulien Roos

Biomechanics Research Group

Sport and Exercise Science Research Group

School for Health

University of Bath

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Appendix A3 GP information sheet younger adults
Movement strategies of younger people to recover from a trip

GP INFORMATION SHEET

Date: 2005 Version: 3

Concerning:

Patient:

Address:

Dear Sir/Madam,

Your patient has been invited and volunteered to take part in a research study about the movement strategies of people (aged 18-35 years) to recover from a trip. We want to inform you about your patient taking part in this experiment. This letter will explain why the research is being done and what it will involve. Please take time to read the following information carefully. Ask us if there is anything that is not clear or if you would like more information. The study has received local research ethics committee approval (project ref. no. 04/Q2001/169).

The purpose of this study is to get better insight into the movement strategies used to recover from a trip. When people get older the base of support during walking increases.

This basically means they make wider steps. This is a more stable manner of walking, but will also slow people down.

In this experiment we want to look at the recovery step length when people recover from a trip. We want to look at the direction and the length of the recovery step. And if so, if this larger recovery step length will cost more energy. What way of recovery is energetically optimal, but also offers good stability? The information we get from this study may be ultimately used as advice for therapies preventing senior people from tripping.

The total duration of this study will be about 6 months, and data collection will take place between May and September 2005. Each participant will be involved for approximately half a day only.

In this study a group of 8 to 10 females between 20 and 35 years old will be studied. Female subjects only were chosen to get a homogeneous group, and because female seniors have been reported to trip more often. The results of these younger subjects will later provide useful comparisons with falling strategies of elderly people.

Your patient has volunteered and been selected to participate in this study. Your patient has completed the Par-Q health screening questionnaire (enclosed) and responded negatively to all standard and additional questions. This means they have met the inclusion criteria set.

What will happen during the experiment?

Your patient will be asked to walk several times (about 24 times) over a walkway (approximately 10 meters). On some occasions obstructing one foot at some part of the walkway will induce a trip. This obstruction will be a hinged bar placed just above the ground. The other foot can move freely to recover from this trip. Your patient will be asked to recover from this trip with a single step. To make sure she cannot fall, she will be secured in a harness fixed overhead. She will be wearing glasses that obscure the lower half of vision to make sure the trip comes as a surprise. An individual with an appropriate work-based first aid qualification will be present during all experiments and all experiments will take place during university medical centre opening hours, which is approximately 400 metres from the laboratory.

To record the movement data the experiments will be video taped. The ground reaction forces from the recovery step will be measured with a force plate. There will also be some markers attached to the body, which will be used for motion analysis.

The results of this study will be used in a three-year research project on movement strategies of senior people to recover from a trip. The results will probably be published in a scientific journal. The patient will not be identified in any publication about this study. After the data have been analysed she will be sent a summary of the results and some personal recommendations based on our findings.

The Bath Local Research Ethics Committee has reviewed this study and granted approval.

If you would like to have more information about this experiment, please contact Paulien Roos (01225 384323).

We will presume you agree with your patient taking part in this experiment if we don't get a response to this letter.

Thank you for reading this.

Paulien Roos

Biomechanics Research Group

Sport and Exercise Science Research Group

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Appendix A4 Participant information sheet older adults

Movement strategies of older people to recover from a trip

PARTICIPANT INFORMATION SHEET

Date: 7 March 2006 Version: 3

Dear participant,

You are being invited to take part in this research study. Before you decide it is important for you to understand why the research is being done and what it will involve. Please take time to read the following information carefully and discuss it with others if you wish. Ask us if there is anything that is not clear or if you would like more information. Take time to decide whether or not you wish to take part.

Injuries from falls becomes an increasing problem with older age, as trips occur more often at older age. The purpose of this study is to get better insight into strategies used to recover from a trip. People with risk of falling increase their base of support when walking. This basically means they make wider steps. This is a more stable manner of walking, but will also slow people down and may cost more energy.

In this experiment we want to look at the recovery step length when people recover from a trip. We want to look at the direction and the length of the recovery step. What way of recovery is energetically optimal, but also offers good stability? An initial experiment has been done with a group of younger participants (20-35 years old). Their results will be compared with those a group of older participants (65-75 years old) participating in this experiment. The information we get from this study may be used as advice for therapies preventing senior people from tripping.

The total duration of this study will be about 6 months, with experiments taking place between January, February and March 2006. Each participant will only need to be involved for two days. The trials on the first day will take about 4 hours, and those on the second day about 6 hours. Rest breaks can be taken at any time during the experiment.

In this study a group of 8 to 10 females between 65 and 75 years old will be studied. Female subjects only were chosen to get a group of people with similar characteristics, and because female seniors have been reported to trip more often. Participants have to be of a maximum weight of 120 kilos (19 stone).

It is up to you to decide whether or not to take part. You will also be asked to sign a consent form. You will still be free to withdraw at any time and without giving a reason. There are some potential risks in this study, an ankle can get strained during a trip for example and there is a small risk of fractures. However the risk of fracture in this study would be the same as when walking along a street in susceptible people. As in ordinary life, it is not possible to exclude all risk.

What will happen to you if you take part?

Initially we will send your GP a letter to inform them that you plan to take part in this experiment. The experiment itself will take two days (4 to 6 hours each day) for each participant. It will take place in the Sport and Exercise Science laboratory of the School for Health at the University of Bath. You will have to sign a written consent form before the experiment will start. After that you will get time to familiarise with the experimental set-up, and you can decide if you are still happy to continue with the experiment. After this you will be asked to walk several times over a walkway (approximately 10 meters (11 yards)). On some occasions (a maximum of 20), the unexpected obstruction of one foot at some part of the walkway will induce a trip. The other foot can move freely to recover from this trip. You will be asked to recover as well as possible from this trip with a single step. To make sure you cannot fall, you will be secured in a full-body safety harness attached to an overhead rail. You will be wearing glasses that obstruct you from seeing the ground to make sure the trip comes as a surprise.

To be able to analyse your motion, anthropometric measurements will be taken. These are measurements of body dimensions. You will lay down on a table and body lengths, widths, depths and perimeters will be measured. We will also measure your body weight using standard scales.

To record the movement data the experiments will be video taped. There will also be some markers attached to the body with tape or to extension attached to your limbs with Velcro bands. These markers will be used to analyse your movements. The application of the markers will require you to be wearing short pants and a sleeveless t-shirt. You will have to wear ankle support to reduce the risk of injury. This support will be provided to you.

All information which is collected about you during the course of the research will be kept strictly confidential. All videotapes will be kept in a locked room, which can only be accessed by the investigators. All computer data will be made anonymous. Any information about you which leaves the department will have your name and address removed so that you cannot be recognised from it.

The results of this study will be used in a three-year research PhD project on movement strategies of senior people to recover from a trip. The results will probably be published in a scientific journal. You will not be identified in any publication about this study. After the data have been analysed you will be sent a summary of the results and some personal recommendations based on our findings.

The Bath Local Research Ethics Committee has approved this study. This proposal is covered by the general compensation arrangements of the University of Bath (contact: Lisa Pritchard, University Insurance Officer, tel: 01225 6378).

If you would like to have more information about this experiment, please contact Paulien Roos (01225 384323), or Grant Trewartha (01225 383055).

Thank you for reading this,

Paulien Roos

Biomechanics Research Group

Sport and Exercise Science Research Group

School for Health

University of Bath

Appendix A5 GP information sheet older adults

Movement strategies of older adults to recover from a trip

GP INFORMATION SHEET

Date: 18 August 2005 Version: 1

Concerning:

Patient:

Address:

Dear Sir/Madam,

Your patient has been invited and volunteered to take part in a research study about the movement strategies of people (aged 65-75 years) to recover from a trip. We want to inform you about your patient taking part in this experiment. This letter will explain why the research is being done and what it will involve. Please take time to read the following information carefully. Ask us if there is anything that is not clear or if you would like more information. The study has received local research ethics committee approval (project ref. no. 04/Q2001/169).

Injuries from falls becomes an increasing problem with older age, as trips occur more often at older age. The purpose of this study is to get better insight into the movement strategies used to recover from a trip. When people get older the base of support during walking increases. This basically means they make wider steps. This is a more stable manner of walking, but will also slow people down.

In this experiment we want to look at the recovery step length when people recover from a trip. We want to look at the direction and the length of the recovery step and whether larger recovery steps require more muscular effort. What way of recovery is energetically optimal, but also offers good stability? An initial experiment has been done with a group of younger participants (20-35 years old). Their results will be compared with those of a group of older participants (65-75 years old) participating in this experiment. The information we get from this study may be ultimately used as advice for therapies preventing senior people from tripping.

The total duration of this study will be about 6 months, and data collection will take place between November and December 2005. Each participant will be involved for two days only. The trials on the first day will take about 4 hours, and those on the second day about 6 hours.

In this study a group of 8 to 10 females between 65 and 75 years old will be studied. Female subjects only were chosen to get a homogeneous group, and because female seniors have been reported to trip more often. The results of these older subjects will be compared with recovery strategies of younger people from a previous similar experiment.

Your patient has volunteered and been selected to participate in this study. Your patient has completed the Par-Q health screening questionnaire (enclosed) and responded negatively to all standard and additional questions. This means they have met the inclusion criteria set.

What will happen during the experiment?

Your patient will be asked to walk several times (about 24 times) over a walkway (approximately 10 meters). On some occasions obstructing one foot at some part of the gait cycle will induce a trip. This obstruction will be a hinged bar placed just above the ground. The other foot can move freely to recover from this trip. Your patient will be asked to recover from this trip with a single step if possible. To make sure she cannot fall, she will be secured in a harness fixed overhead. She will be wearing toe protection and ankle support. She will be wearing glasses that obscure the lower half of vision to make sure the trip comes as a surprise. An individual with an appropriate work-based first aid qualification will be present during all experiments and all experiments will take place during university medical centre opening hours, which is approximately 400 metres from the laboratory.

To record the movement data the experiments will be video taped. The ground reaction forces from the recovery step will be measured with a force plate. There will also be some markers attached to the body, which will be used for motion analysis.

The results of this study will be used in a three-year research project on movement strategies of senior people to recover from a trip. The results will probably be published in a scientific journal. The patient will not be identified in any publication about this study. After the data have been analysed she will be sent a summary of the results and some personal recommendations based on our findings.

The Bath Local Research Ethics Committee has reviewed this study and granted approval.

If you would like to have more information about this experiment, please contact Paulien Roos (01225 384323).

We will presume you agree with your patient taking part in this experiment if we don't get a response to this letter.

Thank you for reading this.

Paulien Roos

Biomechanics Research Group

Sport and Exercise Science Research Group

School for Health

University of Bath

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Appendix A6 Informed consent form

PARTICIPANT CONSENT FORM

Title of Project: Movement strategies of younger people to recover from a trip

Name of Researcher: Paulien Roos

Date: 12 July 2005 Version: 4

Please initial box

1. I confirm that I have read and understand the information sheet dated 12 July 2005 (version 4) for the above study and have had the opportunity to ask questions. ☐

2. I understand that my participation is voluntary and that I am free to withdraw at any time, without giving any reason, without my medical care or legal rights being affected. ☐

3. I give permission to contact my GP about my participation in this experiment. ☐

4. I agree to take part in the above study. ☐

__________________ __________________ __________________

Name of Participant Date Signature

__________________ __________________ __________________

Researcher Date Signature

I consent / do not consent* to the video recordings of my performances to be used in presentations relating to this project on the understanding that at all times my anonymity will be preserved.

* delete as appropriate

____________________	____________________	____________________
Name of subject	Date	Signature
____________________	____________________	____________________
Researcher	Date	Signature

Details of GP:

Name: ____

Telephone number: ____

Address: ____

____

Postcode: ____

Appendix A7 PAR-Q questionnaire

Physical Activity Readiness Questionnaire - PAR-Q (revised 2002)

PAR-Q & YOU

(A Questionnaire for People Aged 15 to 69)

Regular physical activity is fun and healthy, and increasingly more people are starting to become more active every day. Being more active is very safe for most people. However, some people should check with their doctor before they start becoming much more physically active.

If you are planning to become much more physically active than you are now, start by answering the seven questions in the box below. If you are between the ages of 15 and 69, the PAR-Q will tell you if you should check with your doctor before you start. If you are over 69 years of age, and you are not used to being very active, check with your doctor.

Common sense is your best guide when you answer these questions. Please read the questions carefully and answer each one honestly: check YES or NO.

YES	NO		
☐	☐	1.	**Has your doctor ever said that you have a heart condition and that you should only do physical activity recommended by a doctor?**
☐	☐	2.	**Do you feel pain in your chest when you do physical activity?**
☐	☐	3.	**In the past month, have you had chest pain when you were not doing physical activity?**
☐	☐	4.	**Do you lose your balance because of dizziness or do you ever lose consciousness?**
☐	☐	5.	**Do you have a bone or joint problem (for example, back, knee or hip) that could be made worse by a change in your physical activity?**
☐	☐	6.	**Is your doctor currently prescribing drugs (for example, water pills) for your blood pressure or heart condition?**
☐	☐	7.	**Do you know of any other reason why you should not do physical activity?**

If you answered

YES to one or more questions

Talk with your doctor by phone or in person BEFORE you start becoming much more physically active or BEFORE you have a fitness appraisal. Tell your doctor about the PAR-Q and which questions you answered YES.

- You may be able to do any activity you want — as long as you start slowly and build up gradually. Or, you may need to restrict your activities to those which are safe for you. Talk with your doctor about the kinds of activities you wish to participate in and follow his/her advice.
- Find out which community programs are safe and helpful for you.

NO to all questions

If you answered NO honestly to all PAR-Q questions, you can be reasonably sure that you can:

- start becoming much more physically active – begin slowly and build up gradually. This is the safest and easiest way to go.
- take part in a fitness appraisal – this is an excellent way to determine your basic fitness so that you can plan the best way for you to live actively. It is also highly recommended that you have your blood pressure evaluated. If your reading is over 144/94, talk with your doctor before you start becoming much more physically active.

DELAY BECOMING MUCH MORE ACTIVE:

- if you are not feeling well because of a temporary illness such as a cold or a fever – wait until you feel better; or
- if you are or may be pregnant – talk to your doctor before you start becoming more active.

PLEASE NOTE: If your health changes so that you then answer YES to any of the above questions, tell your fitness or health professional. Ask whether you should change your physical activity plan.

Informed Use of the PAR-Q: The Canadian Society for Exercise Physiology, Health Canada, and their agents assume no liability for persons who undertake physical activity, and if in doubt after completing this questionnaire, consult your doctor prior to physical activity.

No changes permitted. You are encouraged to photocopy the PAR-Q but only if you use the entire form.

NOTE: If the PAR-Q is being given to a person before he or she participates in a physical activity program or a fitness appraisal, this section may be used for legal or administrative purposes.

"I have read, understood and completed this questionnaire. Any questions I had were answered to my full satisfaction."

NAME ______________________

SIGNATURE ______________________ DATE ______________________

SIGNATURE OF PARENT or GUARDIAN (for participants under the age of majority) ______________________ WITNESS ______________________

Note: This physical activity clearance is valid for a maximum of 12 months from the date it is completed and becomes invalid if your condition changes so that you would answer YES to any of the seven questions.

continued on other side...

...continued from other side

PAR-Q & YOU

Physical Activity Readiness Questionnaire - PAR-Q (revised 2002)

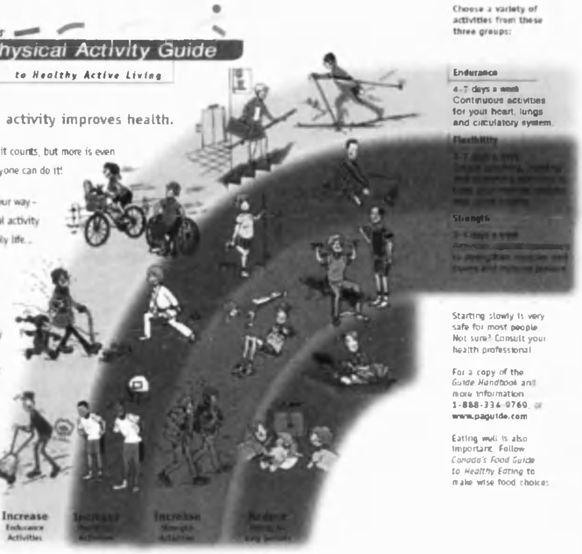
Canada's Physical Activity Guide to Healthy Active Living

Physical activity improves health.

Every little bit counts, but more is even better – everyone can do it!

Get active your way – build physical activity into your daily life...
- at home
- at school
- at work
- at play
- on the way

...that's active living!



Choose a variety of activities from these three groups:

Endurance
4–7 days a week
Continuous activities for your heart, lungs and circulatory system.

Flexibility

Strength

Starting slowly is very safe for most people. Not sure? Consult your health professional.

For a copy of the *Guide Handbook* and more information: **1-888-334-9769**, or **www.paguide.com**

Eating well is also important. Follow *Canada's Food Guide to Healthy Eating* to make wise food choices.

Get Active Your Way, Every Day–For Life!

Scientists say accumulate 60 minutes of physical activity every day to stay healthy or improve your health. As you progress to moderate activities you can cut down to 30 minutes, 4 days a week. Add up your activities in periods of at least 10 minutes each. Start slowly... and build up.

Time needed depends on effort

Very Light Effort	Light Effort 60 minutes	Moderate Effort 30-60 minutes	Vigorous Effort 20-30 minutes	Maximum Effort
• Strolling • Dusting	• Light walking • Volleyball • Easy gardening • Stretching	• Brisk walking • Biking • Raking leaves • Swimming • Dancing • Water aerobics	• Aerobics • Jogging • Hockey • Basketball • Fast swimming • Fast dancing	• Sprinting • Racing

Range needed to stay healthy

You Can Do It – Getting started is easier than you think

Physical activity doesn't have to be very hard. Build physical activities into your daily routine.

- Walk whenever you can – get off the bus early, use the stairs instead of the elevator.
- Reduce inactivity for long periods, like watching TV.
- Get up from the couch and stretch and bend for a few minutes every hour.
- Play actively with your kids.
- Choose to walk, wheel or cycle for short trips.
- Start with a 10 minute walk – gradually increase the time.
- Find out about walking and cycling paths nearby and use them.
- Observe a physical activity class to see if you want to try it.
- Try one class to start – you don't have to make a long-term commitment.
- Do the activities you are doing now, more often.

Benefits of regular activity:

- better health
- improved fitness
- better posture and balance
- better self-esteem
- weight control
- stronger muscles and bones
- feeling more energetic
- relaxation and reduced stress
- continued independent living in later life

Health risks of inactivity:

- premature death
- heart disease
- obesity
- high blood pressure
- adult-onset diabetes
- osteoporosis
- stroke
- depression
- colon cancer

Health Canada Santé Canada · Canadian Society for Exercise Physiology

Source: Canada's Physical Activity Guide to Healthy Active Living, Health Canada, 1998 http://www.hc-sc.gc.ca/hppb/paguide/pdf/guideEng.pdf

FITNESS AND HEALTH PROFESSIONALS MAY BE INTERESTED IN THE INFORMATION BELOW:

The following companion forms are available for doctors' use by contacting the Canadian Society for Exercise Physiology (address below):

The **Physical Activity Readiness Medical Examination (PARmed-X)** – to be used by doctors with people who answer YES to one or more questions on the PAR-Q.

The **Physical Activity Readiness Medical Examination for Pregnancy (PARmed-X for Pregnancy)** – to be used by doctors with pregnant patients who wish to become more active.

References:

Arraix, G.A., Wigle, D.T., Mao, Y. (1992). Risk Assessment of Physical Activity and Physical Fitness in the Canada Health Survey Follow-Up Study. **J. Clin. Epidemiol.** 45:4 419-428.

Mottola, M., Wolfe, L.A. (1994). Active Living and Pregnancy. In: A. Quinney, L. Gauvin, T. Wall (eds.), **Toward Active Living: Proceedings of the International Conference on Physical Activity, Fitness and Health**. Champaign, IL: Human Kinetics.

PAR-Q Validation Report, British Columbia Ministry of Health, 1978.

Thomas, S., Reading, J., Shephard, R.J. (1992). Revision of the Physical Activity Readiness Questionnaire (PAR-Q). **Can. J. Spt. Sci.** 17:4 338-345.

For more information, please contact the:

Canadian Society for Exercise Physiology
202-185 Somerset Street West
Ottawa, ON K2P 0J2
Tel. 1-877-651-3755 • FAX (613) 234-3565
Online: www.csep.ca

The original PAR-Q was developed by the British Columbia Ministry of Health. It has been revised by an Expert Advisory Committee of the Canadian Society for Exercise Physiology chaired by Dr. N. Gledhill (2002).

Disponible en français sous le titre «Questionnaire sur l'aptitude à l'activité physique - Q-AAP (revisé 2002)».

Supported by: Health Canada Santé Canada

Extra information about the PAR-Q

We use this PAR-Q questionnaire for our participants because it covers almost all the medical questions we need to know before we start our experiment. This PAR-Q questionnaire is directed to people aged from 15-69. We will use it also for people older than 69 because our experiment doesn't involve intensive/strenuous exercise, and all participants will be recreationally active.

Supplementary Questions for use with the PAR-Q (University of Bath)

Do you currently smoke? Yes ☐ No ☐

Has your doctor ever said that you have a heart

Arrhythmia? Yes ☐ No ☐

3. Do you have a history of falls? Yes ☐ No ☐

4. Are you presently a participant or have you

   participated in the last month in any research study? Yes ☐ No ☐

Appendix A8 SAFFE questionnaire

SURVEY OF ACTIVITIES AND FEAR OF FALLING IN THE ELDERLY (SAFFE)

Margie E. Lachman, Brandeis University

and

Jonathan Howland, Boston University

Research supported by NIA Roybal Center AG11669

Dear SAFFE User:

As you requested, I am sending a copy of the SAFFE. The scoring information is also included.

I grant you permission to use the SAFFE in your research. Please cite the following reference in your work:

> Lachman, M. E., Howland, J., Tennstedt, S., Jette, A., Assman, S., & Peterson, E. (1998). Fear of Falling and Activity Restriction: The Survey of Activities and Fear of Falling in the Elderly. *Journal of Gerontology: Psychological Sciences, 53B*, P43-P50.

I ask that you please send me preprints and/or reprints of any articles that you prepare which report results with the SAFFE. I am most interested to hear about the research you are doing. Good luck with your research project. Feel free to contact me if you have further questions.

Sincerely yours,

Margie E. Lachman, Ph.D.
Professor

Scoring Information for Survey of Activities and Fear of Falling in the Elderly (SAFFE)

A. Activity Level: Scored as the number of activities they do out of 11. No and nonresponse are given a 0 and a yes is given a 1. Count the number of 1's.

B. Fear of Falling: (see page 46 in Lachman et al., 1998) Recode scoring so that low scores mean low fear: 0 = not at all, 3 = very worried. Recode is 4=0, 3=1, 2=2, 1=3. The fear score is computed as the average worry scores across the 11 activities (or across as many of the activities that are done, i.e., if yes to A). Range is 0 to 3.

F. Activity Restriction: Number of activities that are reported as doing less than used to. That is the number of "less than you used to" responses (response 3) to the question, Compared to 5 years ago, would you say that you.... (range is from 0 to 11).

Scoring the reasons for not doing an activity is optional (see page 48 in the 1998 article):

C. Count the "not at all worried" responses to determine the number of activities that are not done due to reasons other than fear of falling.

D. Count the number of yes responses, to determine the number of activities that are not done because of other reasons in addition to fear of falling.

Activity Questionnaire

A. Do you currently:	1. Go to the store? 1. NO ↓ GO TO C 2. YES ↓ GO TO B	2. Prepare simple meals? 1. NO ↓ GO TO C 2. YES ↓ GO TO B
B. When you........, how worried are you that you might fall?	1. Very worried 2. Somewhat worried 3. A little worried, or 4. Not at all worried GO TO F	1. Very worried 2. Somewhat worried 3. A little worried, or 4. Not at all worried GO TO F
C. Do you not [ACTIVITY] because you are......... that you might fall?	1. Very worried 2. Somewhat worried → GO TO D 3. A little worried Or 4. Not at all worried → GO TO E	1. Very worried 2. Somewhat worried → GO TO D 3. A little worried Or 4. Not at all worried → GO TO E
D. Are there other reasons that you do not........	1. NO 2. YES → SPECIFY:_________ _______________ _______________ GO TO F	1. NO 2. YES → SPECIFY:_________ _______________ _______________ GO TO F
E. What are the reasons that you do not........	SPECIFY:_______________ _______________ _______________ GO TO F	SPECIFY:_______________ _______________ _______________ GO TO F
F. Compared to 5 years ago, would you say that you........	1. More than you used to, 2. About the same, or 3. Less than you used to.	1. More than you used to, 2. About the same., or 3. Less than you used to.

A. Do you currently:	3. Take a tub bath? 1. NO ↓ GO TO C 2. YES ↓ GO TO B	4. Get out of bed? 1. NO ↓ GO TO C 2. YES ↓ GO TO B
B. When you........, how worried are you that you might fall?	1. Very worried 2. Somewhat worried 3. A little worried, or 4. Not at all worried GO TO F	1. Very worried 2. Somewhat worried 3. A little worried, or 4. Not at all worried GO TO F
C. Do you not [ACTIVITY] because you are......... that you might fall?	1. Very worried 2. Somewhat worried 3. A little worried Or → GO TO D 4. Not at all worried → GO TO E	1. Very worried 2. Somewhat worried 3. A little worried Or → GO TO D 4. Not at all worried → GO TO E
D. Are there other reasons that you do not........	1. NO 2. YES → SPECIFY:__________ ______________________ ______________________ GO TO F	1. NO 2. YES → SPECIFY:__________ ______________________ ______________________ GO TO F
E. What are the reasons that you do not........	SPECIFY:__________________ __________________________ __________________________ GO TO F	SPECIFY:__________________ __________________________ __________________________ GO TO F
F. Compared to 5 years ago, would you say that you........	1. More than you used to, 2. About the same, or 3. Less than you used to.	1. More than you used to, 2. About the same, or 3. Less than you used to.

A. Do you currently:	5. Take a walk for exercise? 1. NO ↓ GO TO C 2. YES ↓ GO TO B	6. Go out when it is slippery? 1. NO ↓ GO TO C 2. YES ↓ GO TO B
B. When you........, how worried are you that you might fall?	1. Very worried 2. Somewhat worried 3. A little worried, or 4. Not at all worried GO TO F	1. Very worried 2. Somewhat worried 3. A little worried, or 4. Not at all worried GO TO F
C. Do you not [ACTIVITY] because you are......... that you might fall?	1. Very worried 2. Somewhat worried → GO TO D 3. A little worried Or 4. Not at all worried → GO TO E	1. Very worried 2. Somewhat worried → GO TO D 3. A little worried Or 4. Not at all worried → GO TO E
D. Are there other reasons that you do not........	1. NO 2. YES → SPECIFY:_________ ______________________ ______________________ GO TO F	1. NO 2. YES → SPECIFY:_________ ______________________ ______________________ GO TO F
E. What are the reasons that you do not........	SPECIFY:________________ ______________________ ______________________ GO TO F	SPECIFY:________________ ______________________ ______________________ GO TO F
F. Compared to 5 years ago, would you say that you.....	1. More than you used to, 2. About the same, or 3. Less than you used to.	1. More than you used to, 2. About the same., or 3. Less than you used to.

A. Do you currently:	7. Visit a friend or relative? 1. NO ↓ GO TO C 2. YES ↓ GO TO B	8. Reach for something over your head? 1. NO ↓ GO TO C 2. YES ↓ GO TO B
B. When you........, how worried are you that you might fall?	1. Very worried 2. Somewhat worried 3. A little worried, or 4. Not at all worried GO TO F	1. Very worried 2. Somewhat worried 3. A little worried, or 4. Not at all worried GO TO F
C. Do you not [ACTIVITY] because you are......... that you might fall?	1. Very worried 2. Somewhat worried → GO TO D 3. A little worried Or 4. Not at all worried → GO TO E	1. Very worried 2. Somewhat worried → GO TO D 3. A little worried Or 4. Not at all worried → GO TO E
D. Are there other reasons that you do not....	1. NO 2. YES → SPECIFY:_________ ______________________ ______________________ GO TO F	1. NO 2. YES → SPECIFY:________ ______________________ ______________________ GO TO F
E. What are the reasons that you do not........	SPECIFY:_________________ ______________________ ______________________ GO TO F	SPECIFY:________________ ______________________ ______________________ GO TO F
F. Compared to 5 years ago, would you say that you.....	1. More than you used to, 2. About the same, or 3. Less than you used to.	1. More than you used to, 2. About the same., or 3. Less than you used to.

A. Do you currently:	9. Go to a place with crowds? 1. NO ↓ GO TO C 2. YES ↓ GO TO B	10. Walk several blocks outside? 1. NO ↓ GO TO C 2. YES ↓ GO TO B
B. When you........, how worried are you that you might fall?	1. Very worried 2. Somewhat worried 3. A little worried, or 4. Not at all worried GO TO F	1. Very worried 2. Somewhat worried 3. A little worried, or 4. Not at all worried GO TO F
C. Do you not [ACTIVITY] because you are......... that you might fall?	1. Very worried 2. Somewhat worried → GO TO D 3. A little worried Or 4. Not at all worried → GO TO E	1. Very worried 2. Somewhat worried → GO TO D 3. A little worried Or 4. Not at all worried → GO TO E
D. Are there other reasons that you do not....	1. NO 2. YES → SPECIFY:________ ____________________ ____________________ GO TO F	1. NO 2. YES → SPECIFY:________ ____________________ ____________________ GO TO F
E. What are the reasons that you do not........	SPECIFY:________________ ____________________ ____________________ GO TO F	SPECIFY:________________ ____________________ ____________________ GO TO F
F. Compared to 5 years ago, would you say that you.....	1. More than you used to, 2. About the same, or 3. Less than you used to.	1. More than you used to, 2. About the same., or 3. Less than you used to.

A. Do you currently:	11. Bend down to get something? 1. NO ↓ GO TO C 2. YES ↓ GO TO B
B. When you........, how worried are you that you might fall?	1. Very worried 2. Somewhat worried 3. A little worried, or 4. Not at all worried GO TO F
C. Do you not [ACTIVITY] because you are......... that you might fall?	1. Very worried 2. Somewhat worried → GO TO D 3. A little worried Or 4. Not at all worried → GO TO E
D. Are there other reasons that you do not....	1. NO 2. YES → SPECIFY:_________ ______________________ ______________________ GO TO F
E. What are the reasons that you do not........	SPECIFY:_________________ __________________________ __________________________ GO TO F
F. Compared to 5 years ago, would you say that you.....	1. More than you used to, 2. About the same, or 3. Less than you used to.

Appendix A9 RNHRD bone densitometry referral form

RNHRD BONE DENSITOMETRY DIRECT REFERRAL - DEXA SCAN

Send via your local Referral Management Centre (RMC)

Clinical Measurement Department, Royal National Hospital for Rheumatic Diseases NHS Foundation Trust, Bath BA1 1RL. Direct line telephone: 01225 473414

Patient Name:		**Referring Doctor:**
NHS No.:		
Address:		Address:
Postcode:	Tel.:	Telephone:
Date of birth:	Sex:	Signature: Date:

Patients must have one or more of these risk factors. Please tick all that apply.

1. **Patients under 40 years: refer to Dr A K Bhalla, Rheumatology clinic**
2. **Patients aged 40-60 years must have one of the following risk factors:**

- Long term oral corticosteroids (more than 3 months)
- Hyperparathyroidism
- Chronic respiratory disease
- Vertebral fracture on x-ray
- Rheumatoid arthritis
- Male hypogonadism
- Thyrotoxicosis
- Osteopenic x-ray Please send copy of report
- Malabsorption disorder i.e. coeliac, colitis, liver disease
- Immobility/paraplegia (MS/stroke/other) - state ______ duration ______
- Low trauma fracture since age 50 - site ______

3. **Patients older than 60 years, any of the above and/or any of the following risk factors:**

- Parental hip fracture
- Recent onset vertebral kyphosis Please send copy of lateral x-ray report
- Low BMI (<19)
- Premature menopause (natural/surgical onset < age 45)

4. **Current osteoporosis drug treatment**

- Alendronate
- Etidronate
- Ibandronate
- Risedronate
- Raloxifene
- HRT
- Testosterone
- Calcitonin
- Strontium ranelate
- Teriparatide
- Calcium
- Vitamin D
- Calcitriol

5. **The result of the DEXA scan will influence my decision to**

- Start treatment
- Stop treatment
- Continue/change

Additional information / other drug treatments.

Appendix B Angular momentum equations

Matlab code used to calculate angular momentum.

```
function H = angularmomentumfunction(I,ws,ms,Ys,Zs,Yc,Zc,wcom)

 %This function calculates the angular momentum of body segments relative to body
%centre of mass. Adding up the angular momentums of all segments gives the total
%angular momentum of the body. A positive momentum is in the clockwise direction.

%I: moment of inertia of the segment

%ws: the angular velocity of the segment around its CM

%ms: the mass of the segment

%Ys: the Y coordinate of the CM of the segment

%Zs: the Z coordinate of the CM of the segment

%Yc: the Y coordinate of the body CM

%Zc: the Z coordinate of the body CM

%wcom: the angular velocity of the CM of the segment around the body CM

 %Degrees to radians conversion of the angular velocities

ws2 = -1*ws/180*pi;             %inverse to make positive angular velocity clockwise

wcom2= -1*wcom/180*pi;


n = size(ws,1);

for i = 1:1:n

   r(i) = ((Zs(i)-Zc(i))^2+(Ys(i)-Yc(i))^2)^(0.5);

   H(i,1) = I*ws2(i)+ms*(r(i)^2)*wcom2(i);

end
```

Appendix C Inverse dynamics equations

Matlab code used for the inverse dynamics calculations:

```
function [Fy Fz Mjoint] = IDfunction(Fyds, Fzds, m, Yjds, Yjpx, Zjds, Zjpx,...
Ycom, Zcom, I,Mds, ay, az, alphaddt)

%This function calculates the joint moments of a joint using inverse dynamics
%equations, it also gives the resultant forces (Fy and Fz) acting on the joint. The input
%variables are:
%  Fyds: the force in the y direction on the joint distal from the joint the moment is
%  calculated of. For the ankle joint this is the GRF with ground contact and 0
% if the foot is in the air.
%  Fzds: same as Fyds, but in the z direction
%  m: the mass of the segment distal to the joint
%  Yjds: the Y coordinate of the joint distal from the joint the moment is calculated
%  of. This will be the toes for the ankle joint.
%  Zjds: the Z coordinate of the joint distal from the joint the moment is calculated
%  of.This will be the toes for the ankle joint.
%  Yjpx: the Y coordinate for the joint the moment is calculated of.
%  Zjpx: the Z coordinate for the joint the moment is calculated of.
% Ycom: the Y coordinate of the CM of the segment distal to the joint the moment
% is calculated of.
% Zcom: the Z coordinate of the CM of the segment distal to the joint the moment
% is calculated of.
% I: the moment of inertia of the segment distal to the joint the moment is
%  calculated of.
```

```
%   Mds: the moment of the joint distal from the joint the moment is calculated of.
%  ay: the acceleration in the y-direction of the CM of the segment distal of the
%  joint the moment is calculated of.
% az: the acceleration in the z-direction of the CM of the segment distal of the
% joint the moment is calculated of.
% alphaddt: the angular acceleration of the segment around its centre of mass.
%       t: the time interval over which the joint moments will be calculated.
n = size(alphaddt,1);
for i = 1:1:n;
   dyds(i) = Ycom(i)-Yjds(i);
   dzds(i) = Zcom(i)-Zjds(i);
   dypx(i) = Yjpx(i)-Ycom(i);
   dzpx(i) = Zjpx(i)-Zcom(i);
   Fy(i,1) = Fyds(i) + m*ay(i);
   Fz(i,1) = Fzds(i) + m*9.81 + m*az(i);
   alphaddtrad(i) = alphaddt(i)*pi/180;        %convert the angular acceleration from
                                                 degrees to radians

   Mjoint(i,1) = Mds(i) + Fy(i)*dzpx(i) - Fz(i)*dypx(i) + Fyds(i)*dzds(i)...
                 -Fzds(i)*dyds(i) + I*alphaddtrad(i);
end
```

Appendix D Autolev code of the trip recovery model

```
%-------------------------------------------------------------------
%File D:\paulien\modelling\TripModel\trip52.al
%A 2D torque driven trip model. Consisting of 2 lower legs, 2 upper legs and a
%pelvis. The CM of the upper body is described by its movement pattern. Ground
%contact is modelled with horizontal en vertical spring-damper systems
%The indicator C at the end of a variable name indicates its a variable of the
%contact leg, the S indicates the variable belongs to the swing leg
%Paulien Roos, September 2006
%-------------------------------------------------------------------
%Physical declarations
Newtonian N
Bodies LegCL, LegCU, LegSL, LegSU, FootC, Foot2C, FootS, Foot2S, UB
Points O, ToeC, ToeS, MetC, MetS, AnkleC, AnkleS, KneeC, KneeS, HipS, HipC
Points MidPelvis, UB1, UB2
Frames Pelvis
%-------------------------------------------------------------------
%Mathematical declarations
%
variables U{9}'
variables QAnkleC', QAnkleS', QKneeC', QKneeS', QHipS', QHipC', QLEGCU'
variables QUB, UBOY, UBOZ
variables AnkleCYDisp, AnkleCZDisp, ToeCYDisp, ToeCZDisp, MetCYDisp, MetCZDisp
variables AnkleSYDisp, AnkleSZDisp, ToeSYDisp, ToeSZDisp, MetSYDisp, MetSZDisp
variables POSYUB', POSZUB'
```

constants AnkleCYDispIni, AnkleCZDispIni, MetCYDispIni, MetCZDispIni

constants AnkleSYDispIni, AnkleSZDispIni, MetSYDispIni, MetSZDispIni

constants ToeCYDispIni, ToeCZDispIni, ToeSYDispIni, ToeSZDispIni

constants CMLegCL, CMLegCU, CMLegSL, CMLegSU, CMUB, CMFootC, CMFoot2C, CMFoot2S

constants CMFootS

variables TorkAnkleC, TorkAnkleS, TorkKneeC, TorkKneeS, TorkHipC, TorkHipS

variables AnkleCTork, AnkleSTork, KneeCtork, KneeStork, HipCtork, HipStork

variables PYKneeC, PYHipC, PYUB, PYHipS, PYKneeS, PYAnkleS, PYAnkleC, PYMetC

variables PYToeC, PYToeS, PZKneeC, PZHipC, PZUB, PZHipS, PZKneeS, PZAnkleS

variables PZAnkleC, PZMetC, PZToeC, PZToeS, PXKneeC, PXHipS, PXHipC, PXKneeS

variables PXAnkleS, PXUB, PXAnkleC, PXMetC, PXToeC, PXToeS, PXMidPelvis

variables PYMidPelvis, PZMidPelvis

constants StiffHeelY,StiffHeelZ,DampHeelY,DampHeelZ

constants StiffMetY,StiffMetZ,DampMetY,DampMetZ

constants StiffToeY,StiffToeZ,DampToeY,DampToeZ

constants G=-9.81

constants QMetC, QMetS

constants PI, Ftrip

constants LFootC, LFoot2C, LFootS, LFoot2S, LLegCL, LLegCU, LLegSL, LLegSU

constants LPelvis, LUB

constants UBcoef1, UBcoef2, angvelQUB

Mass UB=MUB, FootC=MFootC, Foot2C=MFoot2C, FootS=MFootS, Foot2S = MFoot2S

Mass LegCL=MLegCL, LegCU=MLegCU, LegSL=MLegSL, LegSU=MLegSU

Inertia FootC,0,0,IFootC

```
Inertia Foot2C,0,0,IFoot2C

Inertia FootS,0,0,IFootS

Inertia Foot2S,0,0,IFoot2S

Inertia LegCU,0,0,ILegCU

Inertia LegCL,0,0,ILegCL

Inertia LegSU,0,0,ILegSU

Inertia LegSL,0,0,ILegSL

Inertia UB,0,0,IUB

%----------------------------------------------------------------

Pi      = 3.14159265358979323846264338327950288419716939937510582

%geometry relating unit vectors

simprot(Foot2S,FootS,3,QmetS)

simprot(Foot2C,FootC,3,QmetC)

simprot(LegCL,FootC,3,QAnkleC)

simprot(LegSL,FootS,3,QAnkleS)

simprot(LegCU,LegCL,3,QKneeC)

simprot(LegSU,LegSL,3,QKneeS)

simprot(Pelvis,LegCU,3,QHipC)

simprot(Pelvis,LegSU,3,QHipS)

simprot(N,LEGCU,3,QLEGCU)

simprot(UB,Pelvis,3,QUB)

%----------------------------------------------------------------

%Position vectors

P_O_MidPelvis>=POSYUB*N1>+POSZUB*N2>

P_MidPelvis_UB1>=-UBOY*Pelvis1>-UBOZ*Pelvis2>

P_UB1_UB2>=UB2>*LUB
```

P_MidPelvis_HipC>=Pelvis3>*0.5*LPelvis

P_HipC_KneeC>=-LegCU2>*LLegCU

P_KneeC_AnkleC>=-LegCL2>*LLegCL

P_AnkleC_MetC>=-FootC2>*LFootC

P_MetC_ToeC>=-Foot2C2>*LFoot2C

P_MidPelvis_HipS>=-Pelvis3>*0.5*LPelvis

P_HipS_KneeS>=-LegSU2>*LLegSU

P_KneeS_AnkleS>=-LegSL2>*LLegSL

P_AnkleS_MetS>=-FootS2>*LFootS

P_MetS_ToeS>=-Foot2S2>*LFoot2S

P_O_UB1>=P_O_MidPelvis>+P_MidPelvis_UB1>

P_O_UB2>=P_O_UB1>+P_UB1_UB2>

P_O_HipC>=P_O_MidPelvis>+P_MidPelvis_HipC>

P_O_KneeC>=P_O_HipC>+P_HipC_KneeC>

P_O_AnkleC>=P_O_KneeC>+P_KneeC_AnkleC>

P_O_MetC>=P_O_AnkleC>+P_AnkleC_MetC>

P_O_ToeC>=P_O_MetC>+P_MetC_ToeC>

P_O_HipS>=P_O_MidPelvis>+P_MidPelvis_HipS>

P_O_KneeS>=P_O_HipS>+P_HipS_KneeS>

P_O_AnkleS>=P_O_KneeS>+P_KneeS_AnkleS>

P_O_MetS>=P_O_AnkleS>+P_AnkleS_MetS>

P_O_ToeS>=P_O_MetS>+P_MetS_ToeS>

P_O_Foot2CO>=P_O_ToeC>+CMFoot2C*P_ToeC_MetC>

P_O_FootCO>=P_O_MetC>+CMFootC*P_MetC_AnkleC>

```
P_O_LegCLO>=P_O_AnkleC>+CMLegCL*P_AnkleC_KneeC>
P_O_LegCUO>=P_O_KneeC>+CMLegCU*P_KneeC_HipC>
P_O_LegSUO>=P_O_HipS>+CMLegSU*P_HipS_KneeS>
P_O_LegSLO>=P_O_KneeS>+CMLegSL*P_KneeS_AnkleS>
P_O_FootSO>=P_O_ToeS>+CMFootS*P_MetS_AnkleS>
P_O_Foot2SO>=P_O_ToeS>+CMFoot2S*P_ToeS_MetS>
P_O_UBO> = P_O_UB1>+CMUB*P_UB1_UB2>

PYAnkleC=DOT(P_O_AnkleC>,N1>)
PYMetC=DOT(P_O_MetC>,N1>)
PYToeC=DOT(P_O_ToeC>,N1>)
PYKneeC=DOT(P_O_KneeC>,N1>)
PYHipC=DOT(P_O_HipC>,N1>)
PYMidPelvis=DOT(P_O_MidPelvis>,N1>)
PYHipS=DOT(P_O_HipS>,N1>)
PYKneeS=DOT(P_O_KneeS>,N1>)
PYAnkleS=DOT(P_O_AnkleS>,N1>)
PYMetS=DOT(P_O_MetS>,N1>)
PYToeS=DOT(P_O_ToeS>,N1>)
PYUB1=DOT(P_O_UB1>,N1>)
PYUB2=DOT(P_O_UB2>,N1>)

PYFootCO=DOT(P_O_FootCO>,N1>)
PYFoot2CO=DOT(P_O_Foot2CO>,N1>)
PYLegCLO=DOT(P_O_LegCLO>,N1>)
PYLegCUO=DOT(P_O_LegCUO>,N1>)
```

```
PYLegSUO=DOT(P_O_LegSUO>,N1>)
PYLegSLO=DOT(P_O_LegSLO>,N1>)
PYFootSO=DOT(P_O_FootSO>,N1>)
PYFoot2SO=DOT(P_O_Foot2SO>,N1>)
PYUBO=DOT(P_O_UBO>,N1>)

PZAnkleC=DOT(P_O_AnkleC>,N2>)
PZMetC=DOT(P_O_MetC>,N2>)
PZToeC=DOT(P_O_ToeC>,N2>)
PZKneeC=DOT(P_O_KneeC>,N2>)
PZHipC=DOT(P_O_HipC>,N2>)
PZMidPelvis=DOT(P_O_MidPelvis>,N2>)
PZHipS=DOT(P_O_HipS>,N2>)
PZKneeS=DOT(P_O_KneeS>,N2>)
PZAnkleS=DOT(P_O_AnkleS>,N2>)
PZMetS=DOT(P_O_MetS>,N2>)
PZToeS=DOT(P_O_ToeS>,N2>)
PZUB1=DOT(P_O_UB1>,N2>)
PZUB2=DOT(P_O_UB2>,N2>)

PZFootCO=DOT(P_O_FootCO>,N2>)
PZFoot2CO=DOT(P_O_Foot2CO>,N2>)
PZLegCLO=DOT(P_O_LegCLO>,N2>)
PZLegCUO=DOT(P_O_LegCUO>,N2>)
PZLegSUO=DOT(P_O_LegSUO>,N2>)
PZLegSLO=DOT(P_O_LegSLO>,N2>)
```

```
PZFootSO=DOT(P_O_FootSO>,N2>)

PZFoot2SO=DOT(P_O_Foot2SO>,N2>)

PZUBO=DOT(P_O_UBO>,N2>)


PXAnkleC=DOT(P_O_AnkleC>,N3>)

PXMetC=DOT(P_O_MetC>,N3>)

PXToeC=DOT(P_O_ToeC>,N3>)

PXKneeC=DOT(P_O_KneeC>,N3>)

PXHipS=DOT(P_O_HipS>,N3>)

PXHipC=DOT(P_O_HipC>,N3>)

PXKneeS=DOT(P_O_KneeS>,N3>)

PXAnkleS=DOT(P_O_AnkleS>,N3>)

PXMetS=DOT(P_O_MetS>,N3>)

PXToeS=DOT(P_O_ToeS>,N3>)

PXMidPelvis=DOT(P_O_MidPelvis>,N3>)

PXUB1=DOT(P_O_UB1>,N3>)

PXUB2=DOT(P_O_UB2>,N3>)


PXFootCO=DOT(P_O_FootCO>,N3>)

PXFoot2CO=DOT(P_O_Foot2CO>,N3>)

PXLegCLO=DOT(P_O_LegCLO>,N3>)

PXLegCUO=DOT(P_O_LegCUO>,N3>)

PXLegSUO=DOT(P_O_LegSUO>,N3>)

PXLegSLO=DOT(P_O_LegSLO>,N3>)

PXFootSO=DOT(P_O_FootSO>,N3>)

PXFoot2SO=DOT(P_O_Foot2SO>,N3>)
```

```
PXUBO=DOT(P_O_UBO>,N3>)
%-----------------------------------------------------------------
%Kinematical differential equations
POSYUB'=U1
POSZUB'=U2
QAnkleC'=U3
QAnkleS'=U4
QHipC'=U5
QHipS'=U6
QKneeC'=U7
QKneeS'=U8
QLEGCU'=U9

% function for upperbody mass movement relative to pelvis (this function has to
% be changed in the C++ routine!!)
UBOY = T*UBcoef1
UBOZ = T*UBcoef2
QUB = T*angvelQUB
%-----------------------------------------------------------------
%Angular velocities and accelerations
W_LegCL_FootC>=QAnkleC'*FootC3>
W_LegSL_FootS>=QAnkleS'*FootS3>
W_LegCU_LegCL>=QKneeC'*LegCU3>
W_LegSU_LegSL>=QKneeS'*LegSU3>
W_Pelvis_LegCU>=QHipC'*Pelvis3>
W_Pelvis_LegSU>=QHipS'*Pelvis3>
```

```
W_N_LEGCU> = QLEGCU'*N3>

W_Foot2C_FootC>=0>

W_Foot2S_FootS>=0>

ALF_LegCL_FootC>=DT(W_LegCL_FootC>,LegCL)

ALF_LegSL_FootS>=DT(W_LegSL_FootS>,LegSL)

ALF_LegCU_LegCL>=DT(W_LegCU_LegCL>,LegCU)

ALF_LcgSU_LegSL>=DT(W_LegSU_LegSL>,LegSU)

ALF_Pelvis_LegCU>=DT(W_Pelvis_LegCU>,Pelvis)

ALF_Pelvis_LegSU>=DT(W_Pelvis_LegSU>,Pelvis)

ALF_Foot2C_FootC>=DT(W_Foot2C_FootC>,Pelvis)

ALF_Foot2S_FootS>=DT(W_Foot2S_FootS>,Pelvis)

ALF_N_LEGCU> = DT(W_N_LEGCU>,N)

%-----------------------------------------------------------------

%Linear velocities

V_O_N>=0>

V_MidPelvis_N> = U1*N1> + U2*N2>

V2PTS(N,UB,MidPelvis,UB1)

V2PTS(N,UB,UB1,UBO)

V2PTS(N,UB,UB1,UB2)

V2PTS(N,Pelvis,MidPelvis,HipC)

V2PTS(N,LegCU,HipC,LegCUO)

V2PTS(N,LegCU,HipC,KneeC)

V2PTS(N,LegCL,KneeC,LegCLO)

V2PTS(N,LegCL,KneeC,AnkleC)

V2PTS(N,FootC,AnkleC,FootCO)

V2PTS(N,FootC,AnkleC,MetC)
```

```
V2PTS(N,Foot2C,MetC,Foot2CO)

V2PTS(N,Foot2C,MetC,ToeC)

V2PTS(N,Pelvis,MidPelvis,HipS)

V2PTS(N,LegSU,HipS,LegSUO)

V2PTS(N,LegSU,HipS,KneeS)

V2PTS(N,LegSL,KneeS,LegSLO)

V2PTS(N,LegSL,KneeS,AnkleS)

V2PTS(N,FootS,AnkleS,FootSO)

V2PTS(N,FootS,AnkleS,MetS)

V2PTS(N,Foot2S,MetS,Foot2SO)

V2PTS(N,Foot2S,MetS,ToeS)


VOFootCOX=DOT(V_FootCO_N>,N3>)

VOFootCOY=DOT(V_FootCO_N>,N1>)

VOFootCOZ=DOT(V_FootCO_N>,N2>)

VOFootSOX=DOT(V_FootSO_N>,N3>)

VOFootSOY=DOT(V_FootSO_N>,N1>)

VOFootSOZ=DOT(V_Foot2SO_N>,N2>)

VOFoot2COX=DOT(V_Foot2CO_N>,N3>)

VOFoot2COY=DOT(V_Foot2CO_N>,N1>)

VOFoot2COZ=DOT(V_Foot2CO_N>,N2>)

VOFoot2SOX=DOT(V_Foot2SO_N>,N3>)

VOFoot2SOY=DOT(V_Foot2SO_N>,N1>)

VOLegCLOX=DOT(V_LegCLO_N>,N3>)

VOLegCLOY=DOT(V_LegCLO_N>,N1>)

VOLegCLOZ=DOT(V_LegCLO_N>,N2>)
```

```
VOLegCUOX=DOT(V_LegCUO_N>,N3>)

VOLegCUOY=DOT(V_LegCUO_N>,N1>)

VOLegCUOZ=DOT(V_LegCUO_N>,N2>)

VOLegSLOX=DOT(V_LegSLO_N>,N3>)

VOLegSLOY=DOT(V_LegSLO_N>,N1>)

VOLegSLOZ=DOT(V_LegSLO_N>,N2>)

VOLegSUOX=DOT(V_LegSUO_N>,N3>)

VOLegSUOY=DOT(V_LegSUO_N>,N1>)

VOLegSUOZ=DOT(V_LegSUO_N>,N2>)

VOMidPelvisX=DOT(V_MidPelvis_N>,N3>)

VOMidPelvisY=DOT(V_MidPelvis_N>,N1>)

VOMidPelvisZ=DOT(V_MidPelvis_N>,N2>)

VOToeCX=DOT(V_ToeC_N>,N3>)

VOToeCY=DOT(V_ToeC_N>,N1>)

VOToeCZ=DOT(V_ToeC_N>,N2>)

VOMetCX=DOT(V_MetC_N>,N3>)

VOMetCY=DOT(V_MetC_N>,N1>)

VOMetCZ=DOT(V_MetC_N>,N2>)

VOToeSX=DOT(V_ToeS_N>,N3>)

VOToeSY=DOT(V_ToeS_N>,N1>)

VOToeSZ=DOT(V_ToeS_N>,N2>)

VOMetSX=DOT(V_MetS_N>,N3>)

VOMetSY=DOT(V_MetS_N>,N1>)

VOMetSZ=DOT(V_MetS_N>,N2>)

VOAnkleCX=DOT(V_AnkleC_N>,N3>)

VOAnkleCY=DOT(V_AnkleC_N>,N1>)
```

VOAnkleCZ=DOT(V_AnkleC_N>,N2>)

VOAnkleSX=DOT(V_AnkleS_N>,N3>)

VOAnkleSY=DOT(V_AnkleS_N>,N1>)

VOAnkleSZ=DOT(V_AnkleS_N>,N2>)

VOKneeCX=DOT(V_KneeC_N>,N3>)

VOKneeCY=DOT(V_KneeC_N>,N1>)

VOKneeCZ=DOT(V_KneeC_N>,N2>)

VOKneeSX=DOT(V_KneeS_N>,N3>)

VOKneeSY=DOT(V_KneeS_N>,N1>)

VOKneeSZ=DOT(V_KneeS_N>,N2>)

VOHipCX=DOT(V_HipC_N>,N3>)

VOHipCY=DOT(V_HipC_N>,N1>)

VOHipCZ=DOT(V_HipC_N>,N2>)

VOHipSX=DOT(V_HipS_N>,N3>)

VOHipSY=DOT(V_HipS_N>,N1>)

VOHipSZ=DOT(V_HipS_N>,N2>)

VOUB1X=DOT(V_UB1_N>,N3>)

VOUB1Y=DOT(V_UB1_N>,N1>)

VOUB1Z=DOT(V_UB1_N>,N2>)

VOUB2X=DOT(V_UB2_N>,N3>)

VOUB2Y=DOT(V_UB2_N>,N1>)

VOUB2Z=DOT(V_UB2_N>,N2>)

VOUBOX=DOT(V_UBO_N>,N3>)

VOUBOY=DOT(V_UBO_N>,N1>)

VOUBOZ=DOT(V_UBO_N>,N2>)

%Linear accelerations

A_O_N>=0>

A_ToeC_N>=DT(V_ToeC_N>,N)

A_MetC_N>=DT(V_MetC_N>,N)

A_ToeS_N>=DT(V_ToeS_N>,N)

A_MetS_N>=DT(V_MetS_N>,N)

A_AnkleC_N>=DT(V_AnkleC_N>,N)

A_AnkleS_N>=DT(V_AnkleS_N>,N)

A_FootCO_N>=DT(V_FootCO_N>,N)

A_FootSO_N>=DT(V_FootSO_N>,N)

A_Foot2CO_N>=DT(V_Foot2CO_N>,N)

A_Foot2SO_N>=DT(V_Foot2SO_N>,N)

A_LegCLO_N>=DT(V_LegCLO_N>,N)

A_LegCUO_N>=DT(V_LegCUO_N>,N)

A_LegSLO_N>=DT(V_LegSLO_N>,N)

A_LegSUO_N>=DT(V_LegSUO_N>,N)

A_MidPelvis_N>=DT(V_MidPelvis_N>,N)

A_KneeC_N>=DT(V_KneeC_N>,N)

A_KneeS_N>=DT(V_KneeS_N>,N)

A_HipC_N>=DT(V_HipC_N>,N)

A_HipS_N>=DT(V_HipS_N>,N)

A_UB1_N>=DT(V_UB1_N>,N)

A_UB2_N>=DT(V_UB2_N>,N)

A_UBO_N>=DT(V_UBO_N>,N)

%%---

%Generalised forces (gravity, extensor torques)

```
TorkAnkleC=AnkleCTork*T
TorkAnkleS=AnkleSTork*T
TorkKneeC=KneeCtork*T
TorkKneeS=KneeStork*T
TorkHipC=HipCtork*T
TorkHipS=HipStork*T
%Joint powers
PowAnkleC=TorkAnkleC*U1'
PowAnkleS=TorkAnkleS*U2'
PowKneeC=TorkKneeC*U3'
PowKneeS=TorkKneeS*U4'
PowHipC=TorkHipC*U5'
PowHipS=TorkHipS*U6'
%Angular and linear momentum
AngMom>=momentum(angular,O)
ZAngMom=DOT(AngMom>,N2>)
XAngMom=DOT(AngMom>,N3>)
LinMom>=momentum(linear)
YMom=DOT(LinMom>,N1>)
ZMom=DOT(LinMom>,N2>)
%Forces
GRAVITY(G*N2>)
TORQUE(FootC/LegCL,TorkAnkleC*LegCL3>)
TORQUE(FootS/LegSL,TorkAnkleS*LegSL3>)
TORQUE(LegCL/LegCU,TorkKneeC*LegCU3>)
TORQUE(LegCU/Pelvis,TorkHipC*Pelvis3>)
```

```
TORQUE(LegSU/Pelvis,TorkHipS*Pelvis3>)
TORQUE(LegSL/LegSU,TorkKneeS*LegSU3>)
%Ground contact forces
AnkleCYDisp = AnkleCYDispIni-PYAnkleC
RYCa=-StiffHeelY*AnkleCYDisp-DampHeelY*VOAnkleCY
AnkleCZDisp = AnkleCZDispIni-PZAnkleC
RZCa=-StiffHeelZ*AnkleCZDisp-DampHeelZ*VOAnkleCZ
FORCE(AnkleC,RYCa*N1>+RZCa*N2>)
FYAnkleC=DOT(Force_AnkleC>,N1>)
FZAnkleC=DOT(Force_AnkleC>,N2>)
MetCYDisp = MetCYDispIni-PYMetC
RYCm=-StiffMetY*MetCYDisp-DampMetY*VOMetCY
MetCZDisp = MetCZDispIni-PZMetC
RZCm=-StiffMetZ*MetCZDisp-DampMetZ*VOMetCZ
FORCE(MetC,RYCm*N1>+RZCm*N2>)
FYMetC=DOT(Force_MetC>,N1>)
FZMetC=DOT(Force_MetC>,N2>)
ToeCYDisp = ToeCYDispIni-PYToeC
RYCt=-StiffToeY*ToeCYDisp-DampToeY*VOToeCY
ToeCZDisp = ToeCZDispIni-PZToeC
RZCt=-StiffToeZ*ToeCZDisp-DampToeZ*VOToeCZ
Force(ToeC,RYCt*N1>+RZCt*N2>)
FYToeC=DOT(Force_ToeC>,N1>)
FZToeC=DOT(Force_ToeC>,N2>)
AnkleSYDisp = AnkleSYDispIni-PYAnkleS
RYSa=-StiffHeelY*AnkleSYDisp-DampHeelY*VOAnkleSY
```

```
AnkleSZDisp = AnkleSZDispIni-PZAnkleS
RZSa=-StiffHeelZ*AnkleSZDisp-DampHeelZ*VOAnkleSZ
FORCE(AnkleS,RYSa*N1>+RZSa*N2>)
FYAnkleS=DOT(Force_AnkleS>,N1>)
FZAnkleS=DOT(Force_AnkleS>,N2>)
MetSYDisp = MetSYDispIni-PYMetS
RYSm=-StiffMetY*MetSYDisp-DampMetY*VOMetSY
MetSZDisp = MetSZDispIni-PZMetS
RZSm=-StiffMetZ*MetSZDisp-DampMetZ*VOMetSZ
FORCE(MetS,RYSm*N1>+RZSm*N2>)
FYMetS=DOT(Force_MetS>,N1>)
FZMetS=DOT(Force_MetS>,N2>)
ToeSYDisp = ToeSYDispIni-PYToeS
RYSt=-StiffToeY*ToeSYDisp-DampToeY*VOToeSY
ToeSZDisp = ToeSZDispIni-PZToeS
RZSt=-StiffToeZ*ToeSZDisp-DampToeZ*VOToeSZ
RYtrip = Ftrip
RZtrip = Ftrip
Force(ToeS,RYtrip*N1>+RYSt*N1>+RZtrip*N2>+RZSt*N2>)
FYToeS=DOT(Force_ToeS>,N1>)
FZToeS=DOT(Force_ToeS>,N2>)
%--------------------------------------------------------------------
%DYNAMICAL DIFFERENTIAL EQUATIONS
ZERO=FR()+FRSTAR()
KANE()
%%--------------------------------------------------------------------
```

```
%%INPUT CONSTANTS, VARIABLES ETC FOR C++ PROGRAM
%%INPUT CONSTANTS, VARIABLES, OUTPUT QUANTITIES
INPUT TINITIAL=0.0, TFINAL=2.0, INTEGSTP=0.001, PRINTINT=5
INPUT ABSERR=1.0E-08, RELERR=1.0E-07
%----------------------------------------------------------------
%UNITS FOR T,CONSTANTS,VARIABLES,OUTPUT QUANTITIES
UNITS [CMLegCL,CMLegCU,CMLegSL,CMLegSU,CMFootC,CMFoot2C, CMFootS, CMFoot2S]=NONE
UNITS [QAnkleC,QKneeC,QHipS,QHipC,QKneeS,QAnkleS,QLEGCU]=DEG
UNITS [AnkleCTork,AnkleSTork,HipCTork,HipSTork,KneeCTork,KneeSTork] =N*m
UNITS [U1,U2,U3,U4,U5,U6,U7]=DEG/S
UNITS [U8,U9]=M/s
UNITS [LLegCL,LLegCU,LLegSL,LLegSU,LFootC,LFoot2C,LFootS,LFoot2S, LPelvis] =M
UNITS [AnkleCYDisp,AnkleCZDisp,MetCYDisp,MetCZDisp,ToeCYDisp, ToeCZDisp] =M
UNITS [AnkleSYDisp,AnkleSZDisp,MetSYDisp,MetSZDisp,ToeSYDisp, ToeSZDisp] =M
UNITS T=SEC
UNITS [MLegCL,MLegCU,MLegSL,MLegSU,MUB,MFootC,MFoot2C,MFootS, MFoot2S]=kg
UNITS G=M/SEC^2
UNITS [ILegCL,ILegCU,ILegSL,ILegSU,IFootC,IFoot2C,IFootS,IFoot2S]=kg*m^2
UNITS [AnkleCYDispIni,AnkleCZDispIni,AnkleSYDispIni,AnkleSZDispIni]=M
UNITS [ToeCYDispIni,ToeCZDispIni,ToeSYDispIni,ToeSZDispIni]=M
UNITS [MetCYDispIni,MetCZDispIni,MetSYDispIni,MetSZDispIni]=M
UNITS [DampHeelY,DampHeelZ,DampMetY,DampMetZ,DampToeY,DampToeZ] =N/M
```

```
UNITS [StiffHeelY,StiffHeelZ,StiffMetY,StiffMetZ,StiffToeY,StiffToeZ]=N/M/S

UNITS [POSYUB,POSZUB]=M

%-----------------------------------------------------------------

OUTPUT
T,QAnkleC,U1,U1',QAnkleS,U2,U2',QKneeC,U3,U3',QKneeS,U4,U4',QHipC,U5,U5',QH
ipS,U6,U6',QUB,QLEGCU, U9, U9'

OUTPUT
T,PYToeC,PZToeC,PYToeS,PZToeS,PYAnkleC,PZAnkleC,PYAnkleS,PZAnkleS,PYMet
C,PZMetC,PYMetS,PZMetS,PYKneeC,PZKneeC,PYKneeS,PZKneeS,PYHipC,PZHipC,P
YHipS,PZHipS,PYMidPelvis,PZMidPelvis,PYUB1,PZUB1,PYUB2,PZUB2

OUTPUT
T,PXToeC,PXToeS,PXAnkleC,PXAnkleS,PXMetC,PXMetS,PXKneeC,PXKneeS,PXHip
C,PXHipS,PXMidPelvis,PXUB1,PXUB2

OUTPUT
T,PYFootCO,PZFootCO,PYFoot2CO,PZFoot2CO,PYLegCLO,PZLegCLO,PYLegCUO,P
ZLegCUO,PYLegSUO,PZLegSUO,PYLegSLO,PZLegSLO,PYFootSO,PZFootSO,PYFoo
t2SO,PZFoot2SO,PYUBO,PZUBO

OUTPUT
T,PXFootCO,PXFoot2CO,PXLegCLO,PXLegCUO,PXLegSUO,PXLegSLO,PXFootSO,P
XFoot2SO,PXUBO

OUTPUT
T,AnkleCYDisp,AnkleCZDisp,MetCYDisp,MetCZDisp,ToeCYDisp,ToeCZDisp

OUTPUT T,AnkleSYDisp,AnkleSZDisp,MetSYDisp,MetSZDisp,ToeSYDisp,ToeSZDisp

OUTPUT T,PowAnkleC,PowAnkleS,PowKneeC,PowKneeS,PowHipC,PowHipS

OUTPUT T,ZAngMom,XAngMom,YMom,ZMom

OUTPUT T,FYAnkleC,FZAnkleC,FYMetC,FZMetC,FYToeC,FZToeC

OUTPUT T,FYAnkleS,FZAnkleS,FYMetS,FZMetS,FYToeS,FZToeS

OUTPUT T,VOTOECY, VOTOECZ,VOTOESY, VOTOESZ,
VOFOOTCOY,VOFOOTSOY,VOFOOT2COY,VOFOOT2SOY
```

```
%------------------------------------------------------------------
%WRITE C GENERATION FOR NUMERICAL SOLUTION
CODE DYNAMICS() D:\paulien\modelling\TripModel\trip52.c
SAVE  D:\paulien\modelling\TripModel\trip52.all
```